

STATIC ASYMMETRICAL WHOLE BODY EXERTION IN HUMANS

by

ALEXIS TRACEY WILKINSON

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Human Performance Laboratory
Department of Anatomy and Developmental Biology
Royal Free Hospital School of Medicine
London

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ABSTRACT.

The asymmetrical postures of subjects were recorded simultaneously with the force applied by one hand on a handle at three heights during exertion in specific directions, including many with a left or right directional component. A biomechanical analysis of the exertions was made in the horizontal plane and in the vertical plane containing the force vector. Analysis in the vertical plane showed that subjects achieved greater forces by both increased muscular effort and more effective deployment of body weight.

Analysis in the horizontal plane revealed the existence of a horizontal moment at the foot-base. This moment was found to be small or negligible when a person exerted in the fore-aft plane, but was of considerable magnitude when exertion was carried out in directions with a lateral component.

To investigate the generation of this moment, two further experiments were conducted. The first used surface electromyography to explore the roles of the flexors and extensors of the lower limb, while the second utilized a force plate to examine in detail the forces and moments at the foot-base. It was found that for laterally directed exertions, approximately half of the moment was generated by one foot exerting a force on the floor in a direction opposite to the other, and half was produced by each foot exerting a horizontal torque individually. The quadriceps, hamstrings and tibialis anterior were implicated in the first mechanism.

The horizontal turning moment at the feet has not previously been recognised and should be incorporated into models of asymmetrical exertion.

*Success is going from failure to failure
with undiminished enthusiasm.*

Winston Churchill (1874 - 1965).

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CHAPTER 1.

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INTRODUCTION

1. INTRODUCTION.

During the course of a normal day, a person is frequently required to exert manual forces, although the magnitude of these forces is often small. Most of these exertions are performed in asymmetrical postures, often one-handed (which in itself implies asymmetry), and many are exerted in directions with a lateral component. Everyday activities such as holding open a door, changing gear in a car, or gripping on to a support while standing in a moving train all require forces to be exerted in a lateral direction. In industrial manual materials handling, the amount of force required may be of considerable magnitude, and the body may be subjected to much greater stresses. Pushing a heavy box sideways along a shelf or operating the levers of machinery are examples of tasks that may require substantial lateral forces to be exerted. During these exertions, the demands on the body (and therefore the patterns of muscle activity) must differ from those associated with exertions in the sagittal plane (lifting, pulling and pushing), which have been more commonly investigated by researchers.

How the body is employed to perform one-handed, static, often very asymmetrical postures and exertions while standing has not been comprehensively investigated and it is part of the purpose of this thesis to explore the issue further. The relationships between posture and strength have been considered, but again most of the studies in this area have been concerned with symmetrical tasks, where the direction of forces exerted is in the sagittal plane. Another objective of the thesis is therefore to examine the way in which a person's posture affects his or her

ability to exert forces in directions with a lateral component.

The thesis begins with a literature review in Chapter 2. The literature dealing with the subject of human strength is vast, so topics considered are those relating to the main themes of the thesis. Since it forms the background to this study, static symmetrical whole body exertion is considered first, followed by asymmetrical and, briefly, dynamic exertion. Biomechanical modelling, prediction of strength and factors that limit strength are then reviewed. Because the act of exerting laterally requires the production of torques in the horizontal plane, this topic is then considered. Finally, the use and limitations of electromyography are discussed.

The experimental work is begun in Chapter 3 with a study of relationships between posture and strength. An extensive study investigating the postures of subjects during exertion in many directions, including several outside the fore-aft plane, is presented. Postural analysis, together with measurement of exerted forces, allowed consideration of postures associated with particular directions and magnitudes of forces. A biomechanical analysis provided information regarding the differences between the demands made upon the body when engaged in exerting forces in the fore-aft plane with those when exerting in directions with a lateral component. A notable feature of the latter type of exertion was the presence of a horizontal moment of some magnitude at the foot-base.

Once this important difference between exerting in directions in the fore-aft plane and laterally had been established, a more detailed investigation into the horizontal

moment could be undertaken. In Chapter 4, possible mechanisms are suggested for its generation. Two further experiments to explore these possibilities are then described. The first used surface electromyography to investigate the role of some of the muscles of the lower limb in the generation of the moment. The second utilized a force plate to measure forces and moments at the foot-base, and to investigate the strategies required for the application of both torque and translational forces at the floor simultaneously.

Chapter 5 provides a general discussion, suggestions for future work and the general conclusions of the thesis.

1.1 GENERAL AIMS OF THE THESIS.

1. To investigate relationships between posture and strength during exertion in asymmetrical postures with forces directed outside the fore-aft plane.
2. To determine how the demands on the body during asymmetrical, laterally directed exertions differ from exertions directed in the fore-aft plane.
3. To examine how the additional demands are met.
4. To consider the role of the flexors and extensors of the lower limb during one-handed, laterally directed exertions.

1.2 Ethics clearance for the experiments was obtained from the Defence Research Agency (Centre for Human Sciences). All subjects gave informed consent.

CHAPTER 2.

LITERATURE REVIEW

2.1 ANALYSIS OF STATIC WHOLE BODY EXERTION.

In the search for the causes and prevention of injury (largely back pain) at the workplace, which continues to take its toll on industry in worker compensation and lost working days, a vast body of research has been amassed in the fields of biomechanics, physiology, epidemiology and psychophysics. It is hoped that with a greater understanding of the mechanisms involved in the execution of manual tasks, the incidence of injuries in the work environment may eventually be reduced. Early work concentrated on static, symmetrical exertions in the sagittal plane, such as two-handed lifting, pushing and pulling.

2.1.1 Lifting.

Whitney (1958) measured the maximal lifting forces that subjects exerted on a force bar while standing on a force platform, under various conditions of foot placement, bar height, grasp and type of lift. He developed an equation that related the magnitude of the lifting force to the foot placement, bar height and body weight, and suggested that the maximum lifting force is limited by the counterbalancing moment provided by the body weight. Only when the feet were positioned close to the axis of grasp did inherent muscular strength become an important factor. Maximal forces (approximately 780N) were achieved with the back of the foot 0.3m behind the lifting bar, with lower values (approximately 330N) with the foot 0.5m behind the bar.

2.1.2 Pushing and pulling.

Gaughran and Dempster (1956) used biomechanical principles to analyze forces in a system where subjects sat on a force platform and either pushed or pulled on a fixed handle. They attempted to investigate the effect of the "dead weight" of the body during these exertions by placing the subjects in certain postures. The postures, forces acting on the platform, and applied forces on the handle were simultaneously recorded. Their results indicated that two force couples were balanced in the static equilibrium of the experimental conditions. The first consisted of vertical forces - the weight of the body acting downward, and the reactive force at the "point of effective seat contact" acting upward. The second consisted of horizontal forces - the force (push or pull) at the hands, and the horizontal force on the seat. By careful measurement of the moments, they discovered a linear relationship between the length of the moment arm of the vertical couple and the horizontal force produced at the hands. Adding a footplate at some distance in front of the platform further increased the moment arm between the vertical forces and thus also the horizontal force at the hands. A later experiment was carried out (Dempster, 1958), where a single subject exerted pulling forces on a chain in many different directions while standing with or without a front support at the feet and while seated. The strength of the pull was directly dependent on the deployment of body weight and "muscles at the various joints of the body had no other mechanical significance in effecting pull magnitude than that of making the joints rigid so that a given posture could be maintained".

Although their conclusions were valid for their particular experimental situation, later work by Grieve and Pheasant (1982b) showed that the results could not be applied in all conditions of exertion (see Section 2.1.3, p.25 below).

Ayoub and McDaniel (1974) attempted to define an optimal posture for a person exerting a pushing or pulling force by measuring maximal horizontal forces with varying hand heights and distances of the feet from the load. From this information, they determined the pushing "efficiency" (horizontal force as a percentage of total force applied). Expressing the bar height and distance of load from the feet as a percentage of reach height (distance from floor to grip centre in an overhead reach) of each subject, they calculated that the most "efficient" height of the bar was 50%, and foot position 70%, of reach height. However, the largest magnitudes of horizontal pushing force (up to 670N for males and 330N for females) were achieved at a bar height between 60 and 80%, although with a greater proportion of vertical force. The largest pulling forces (up to 670N for males and 400N for females) were achieved at a lower bar height - between 30 and 50% of reach height.

The "optimum" positions were defined as those which were "efficient, where the operator has the best muscular advantage and are the least stressful", although no attempt to define the latter two terms was made. These were then calculated as the time a subject could maintain a maximum pushing force until a 10% drop off was observed, although no evidence was presented to indicate that there was a relationship between force duration and their definition of an "optimum" position

for bar height and foot position.

They also tested the maximum pushing and pulling force of subjects of similar height (50th percentile), but representative of the 5th, 50th and 95th percentile in weight of the male population. Their results showed that the heavier group of subjects produced consistently greater forces than the lighter in both pushing and pulling at all foot distances and bar heights.

Chaffin *et al.* (1983) studied the postures of subjects pushing and pulling with one or two hands while standing with feet either together or separated (one in front of the other). During pulling with separated feet, the back foot was only slightly further from the handle than the front foot, and there was little change in the exerted force (means 216N and 217N), but pushing forces were substantially increased when the feet were separated and the back foot moved to a position much further from the handle (from mean 227N to 276N). It was suggested that the weight of the forward placed lower limb increased "the turning moment about the foot" and also that the subject felt protected from slipping, although the authors had provided a surface with a high coefficient of friction.

2.1.3 Exertion in all directions in the sagittal plane.

Grieve (1979a; 1979b) examined whole body posture during standing exertions, and introduced the Postural Stability Diagram, which is a graphical representation of the static forces (horizontal and vertical) being applied during force exertion.

This method could be used to express two sets of constraints for any given task - personal or environmental. The personal statements referred to characteristics of the person engaging in the task, such as position of hand centroid and centre of foot pressure, or the maximum strength of exertion in a particular direction (Grieve, 1979a), while environmental statements referred to limitations of the environment, such as the demands of the task or frictional considerations at the floor (Grieve, 1979b). Figure 2.1a gives an example of a Postural Stability Diagram. The inner scales refer to forces at the hands (or point of application of force) with the origin at the centre of the square, and the outer scales refer to forces at the feet (or centre of support), with the origin at the centre-base. Any point on the diagram will then represent the forces at the hands and feet. The vector joining the point to the centre represents the force at the hands, while the line connecting the centre-base to the point represents the force at the feet. The vertical line between the centre and centre-base represents body weight and completes the triangle of vectors relevant for static equilibrium. By taking moments about the centre of foot pressure, the equation of static exertion was developed:

$$\frac{LIFT}{W} = \frac{h}{b} \cdot \frac{PUSH}{W} - \frac{a}{b} - \frac{TWIST}{bW}$$

where h is the height of the hand centroid, a is the horizontal distance between the centre of gravity and the centre of foot pressure, b is the horizontal distance

Figure 2.1: a) Left: person exerting forces and torques at centroid of force at hands (H) and feet (F), with G as centre of gravity. Right: Postural Stability Diagram where static forces at hands and feet can be represented (from Grieve, 1979a). b) Left: manual force in exertion, showing *live* and *dead* axes. Centre: dead-weight log analogy of exertion in *dead* axis. Right: Jack-in-a-Box analogy of exertion in the *live* axis (from Grieve and Pheasant, 1982b).

between the hand centroid and the centre of foot pressure, and W is the body weight. The slope of this line is the same as that of a line connecting the hand centroid to the centre of foot pressure. A maximal force, which is not restricted by physiological or frictional considerations, will have the head of its vector along the line of the equation. The concepts of the Postural Stability Diagram and equation of static exertion were tested and validated experimentally.

The force applied by a subject (Grieve and Pheasant, 1982b) could be divided into *dead* and *live* components (Figure 2.1b). The *live* component of force lies along the line of the equation of static exertion, while the *dead* component is at right-angles to it. The latter depended totally on the body weight and its deployment in space, and was analogous to a block of wood leaning against a handle. The *live* force was compared to the force a Jack-in-a-Box exerts (Figure 2.1b). In a person, this would relate to his or her muscular capacity. Although the block of wood would require no "exertion" to remain in position, a person would require active muscles as well as passive mechanisms to maintain the posture. Since exertion of the *live* force also involves active muscles, the *live* and *dead* forces could not easily be separated in terms of muscular effort. When maximal forces in the sagittal plane were measured experimentally with differing hand and foot positions, it was found that if the force vector lay close to the *dead* axis, 62% of the variance could be explained by variation in the height and weight of the population, but if it were close to the *live* axis, only 22% was explained (Pheasant, 1977).

These ideas helped provide an explanation for the findings of some of the earlier studies above. In the experiments of Gaughran and Dempster (1956), the force applied in all cases was purely horizontal in the sagittal plane. A linear relationship between the length of the moment arm for the vertical couple and the applied horizontal force was found. These equate to the terms a and PUSH respectively in the equation of static exertion. Since TWIST and LIFT were zero, then:

$$h \cdot \frac{PUSH}{W} = a$$

so that $PUSH \propto a$ for a given value of h and W .

Whitney (1958) stated that "when the plane of the grasping axis is coincident with the frontal plane in which the foot pivots lie, then clearly the maximum lifting force is limited entirely by the muscular capacity for body extension". Under these conditions of lift, the *live* component of force is by far the larger. The terms in Whitney's equation can be compared to those in the equation of static exertion, so that $HV = LIFT$, $-HH = PUSH$, $W = W$, $p - \beta = b$, $-\alpha = a$, and $h = h$, where the first terms are from Whitney. Substituting Whitney's terms in the equation of static exertion, gives:

$$\frac{HV}{W} = -\frac{h}{(p-\beta)} \cdot \frac{HH}{W} + \frac{\alpha}{(p-\beta)}$$

From Whitney, $HV \cdot \tan \theta = HH$, where θ is the angle between the resultant force at the hands and its vertical component (HV or LIFT). Rearranging and substituting for HH gives:

$$HV = \frac{W\alpha}{(p+h \cdot \tan\theta - \beta)}$$

which is Whitney's original equation.

Grieve and co-workers carried out a number of experiments with subjects exerting with two hands in various directions in the fore-aft plane to investigate the influence of a variety of personal and environmental constraints on strength, such as foot placement, bar height, the incorporation of an inhibiting wall or ceiling, posture, and friction at the feet (Pheasant and Grieve, 1981; Pheasant *et al.* 1982; Grieve, 1983a; 1983b). A definition of efficiency of exertion was also proposed as a replacement to the suggestion of Ayoub and McDaniel (1974) that pushing efficiency could be measured by expressing the horizontal (desired) force as a percentage of the total force (see section 2.1.2, p.24 above). The new definition explored the possibility that a person can achieve a greater force in a particular direction by exerting in a different direction to that required (the maximum advantage of using a component of exertion - MACE) (Grieve and Pheasant, 1981). A limited study with constrained experimental conditions (foot and hand placement) showed that six naive subjects knew instinctively how to utilize this concept.

Maximal forces in all directions in the fore-aft plane were found to vary depending on handle height and foot placement (Pheasant and Grieve, 1981). The approximate range of maximal forces was as follows: lifting - 100 to 900N (males), 50 to 650N (females); pushing - 150 to 500N (males), 100 to 380N (females); pressing down - 580 to 700N (males), 450 to 500N (females); pulling - 80 to 400N (males), 50 to 300N (females).

Rohmert and Mainzer (1987) also suggested a graphical method of displaying the strength of individuals in a way that could have a more general application than specific cases of strength measurement. The method, called a Vektogramm, shows some similarities to the Postural Stability Diagram of Grieve (1979a; 1979b). A graphical representation of maximal static strength is defined, where the origin in a co-ordinate system equates to the centre of grip, and vectors represent the spatial direction of forces.

2.2 STATIC ASYMMETRICAL EXERTION.

Although some studies have examined exertion in asymmetrical postures, few have investigated static analysis of exertions in directions other than in those in the fore-aft plane and it is part of the intention of this thesis to explore the area further.

Kroemer (1974) investigated maximal pushing forces under a large number of conditions - with one hand, both hands or the back applying horizontal forces either with or without a foot rest or wall to brace against. He showed that body

posture, body support and the chain through which the force vectors flow are important in affecting the ability to exert forces. The lowest forces were exerted with one hand (5th percentile $\pm 250\text{N}$) and the greatest with the back against a low board with the feet braced against a vertical surface (5th percentile $\pm 750\text{N}$). He discovered that being able to brace against a surface substantially increased the amount of pushing force that could be applied, particularly if the surface was close to a position horizontally opposite the level at which the pushing force was applied (e.g., a wall for the feet to push against while applying force with the lower back in a sitting position or for the back to push against while applying force with the hands in a standing position). In these experiments, body weight was not found to be a practical predictor of force capability.

Warwick *et al.* (1980) provided one of the few studies of static asymmetrical exertion out of the fore-aft plane when they tested the maximal voluntary contraction of twenty-nine subjects in symmetric or asymmetric (rotated torso) postures as they pushed, pulled, lifted, pressed down and exerted to the right and to the left, on a handle placed at two heights, with one or both hands. The forces applied by the subjects varied considerably depending on the test conditions (direction of exertion, foot position and hands used), but exerting to the right or left gave consistently lower forces than directions in the sagittal plane with the handle at shoulder height, except for one extreme one-handed posture, where the lifting strength was reduced below that of pushing to the left. The mean forces measured for asymmetrical exertions with one or both hands at shoulder height in the fore-aft plane were: lifting - 155N, pressing down - 178N, pushing - 193N,

pulling - 149N, while the forces directed laterally were: to the right - 98N, to the left - 109N. When the handle was at knee height, the results were less consistent, with the strength of exertion to the left or right frequently exceeding that of lifting. It would appear that at the lower handle height, the difference between forces exerted laterally and those directed upward is less marked, perhaps because the possibility exists of deploying body weight more effectively.

Fothergill *et al.* (1991) expanded some aspects of the studies by Grieve and co-workers (section 2.1.3) by investigating one- and two-handed exertions in all directions in the vertical fore-aft plane. The difference in the magnitude of applied force between these was small in many directions, particularly at higher hand heights (Figure 2.2). They concluded that in these cases, the limiting factor in the strength of the exertions was either a) the deployment of body weight in relation to the centre of foot pressure or b) a part of the body other than the upper limbs. In some cases, the ratio of one- to two-handed strength was greater than one. Having one hand in contact with the handle allowed twisting of the body, and therefore greater mobility of the centre of gravity relative to the foot-base.

Sanchez and Grieve (1992) carried out an extensive study with two groups of nine subjects to provide data on static one- and two-handed, symmetrical and asymmetrical lifting strength in 96 different postures. The hand positions during the exertions included six heights, two reach distances and five planes (degrees of twist). The findings indicated that strength was only weakly dependent on the degree of twist, strongly dependent on the height of the hands and correlated

Figure 2.2: Above: Plots of two-handed maximal exertions in all directions in the fore-aft plane at two handle heights for one subject. Outer circle represents forces equal to 100% body weight.

Below: Ratio of mean one-handed/mean two-handed strength.

Centre = 0, inner circle = 0.5, outer circle = 1. Radiating lines indicate areas where two-handed exertions are significantly ($p < 0.05$) greater than one-handed (from Fothergill *et al.* 1991).

significantly with body weight in almost all postures.

Pinder *et al.* (1995), who measured the one-handed strength of subjects exerting in all directions at three handle heights, found that the weakest applied forces were "those involving large components of sideways forces", but in contrast to Warwick *et al.* (1980), this applied to all three handle heights. In this study, the greatest forces were exerted forward and slightly to the left of the fore-aft plane in pushing and backward and slightly to the right when pulling. Presumably this is a reflection of the fact that the exertions were right-handed.

Researchers who have developed dynamic three-dimensional biomechanical models (see section 2.4, p.40 below) have found that while asymmetric lifting gives greater shear forces on the spine, the compressive forces are smaller than those in symmetric lifting.

2.3 DYNAMIC EXERTION.

More recently, research in manual materials handling has expanded to include studies in dynamic symmetrical and asymmetrical exertion. Industrial studies have tended to use physiological and/or psychophysical methods to analyze strength (Snook, 1978; Kumar, 1980; Garg, 1983; Ciriello and Snook, 1983; Mital and Fard, 1986; Drury *et al.* 1989a), but biomechanical aspects have also been considered (Drury *et al.* 1989b), particularly in the field of modelling (see section 2.4, p.40). The results from one measured parameter often differ from the results of another,

which makes it difficult for investigators to advocate definite recommendations.

Kumar and Davis (1983) attempted to distinguish between static postural loading and dynamic activity in terms of heart rate, electromyography of the erector spinae and intra-abdominal pressure measurements. Subjects lifted or lowered a weight, or were gradually loaded or unloaded while maintaining a static posture. Dynamic postural activity gave higher readings for both electromyography and intra-abdominal pressure, but values for the increase in heart rate were similar under both conditions.

Garg and Badger (1986) measured the maximal isometric strength and dynamic maximal acceptable weight of lift of subjects in both symmetrical and asymmetrical postures. They compared static and dynamic values, finding that static strength accounted for 62% of the variance of the maximal acceptable weight of lift during the dynamic lifts.

Tsuang *et al.* (1992) estimated the peak moments at the hip and L5/S1 using a) a static model, b) a static model incorporating the inertial properties of the load, and c) a dynamic model, while subjects lifted two weights at two speeds, fast and slow. The three methods yielded quite different results. The peak moments were least with the static model, greater with the static + inertia model, and greatest with the dynamic model (87% (L5/S1) and 95% (hip) greater than those calculated with the static model). Leskinen (1985) compared four types of lift using both static and dynamic analyses and found that the peak compression at L5/S1 increased by

between 30 and 60% for the dynamic analyses. These studies demonstrate the difficulty of using static analyses to predict what will occur in a dynamic situation.

In 1981, a committee from the National Institute of Occupational Safety and Health (NIOSH) in the USA attempted to provide industrial guidelines for maximal acceptable weights of lift, based on research from the fields of epidemiology, biomechanics, physiology and psychophysics (NIOSH, 1981). They presented an equation, valid only for symmetrical, sagittal plane lifting, for use in the analysis of industrial tasks. The lift is analyzed in terms of horizontal distance from the centre of the ankles, vertical starting height above the ground, vertical distance the load is moved and frequency of lift. The level of risk of a task could be classified into one of three categories, separated by two limits, the Action Limit (AL) and the Maximum Permissible Limit (MPL). The maximum possible AL is a lift of 40kg, which would be given by the equation under optimal conditions i.e., when the lift is close to the body (0.15m), begins at knuckle height (0.75m), is moved vertically 0.25m or less, and is lifted at a low frequency. A change in any of these factors is thought to be less than optimal and the load is reduced accordingly. A value below that of the AL is deemed acceptable for most workers (over 99% of men and 75% of women), while a value falling between the AL and the MPL is judged to be acceptable only for some workers (under 25% of men and 1% of women) and would require increasing attention to modifications either in worker selection and training or job design. A value above the MPL is judged to be hazardous in terms of risk of injury and complete re-design of the task is recommended (e.g., resort to mechanical lifting aids).

Because of the limited application of the original NIOSH equation and considering the wealth of new research since 1981, a new committee suggested modifications to the equation in 1991 (Waters *et al.* 1993). As well as changes to the terms in the original equation, hand-load coupling and asymmetry of lift were additionally considered for the calculation of a Recommended Weight Limit (RWL), which replaced the AL. The maximum possible RWL is a lift of 23kg under optimal conditions i.e., when the load is lifted close to the body (0.25m), is moved vertically 0.25m or less, begins at knuckle height (0.75m), and is lifted at a low frequency (once every five minutes or less) with a symmetrical posture and a good hand-handle interface. A simple ratio of the load lifted to the RWL then gave a Lifting Index to be used in the analysis of the task in question. A value of the Lifting Index more than one is thought to pose a greater risk for many workers, but even the committee itself seemed divided on this point. The level of risk for any value of the Index is still far from being quantified.

A critical review of the five methods employed in the evaluation of the severity of a manual task was undertaken by Leamon (1994). He reviewed the literature on work physiology, intra-abdominal pressure, strength capacity, biomechanical modelling and psychophysics and tried to relate these to the epidemiology of low back pain, often without much success. He points out that much field validation of many of the assertions made by researchers in these areas still needs to be done. He suggests that the question to be answered is, "What is the significance of following a particular design criterion to the reduction of low back pain disability?" and adds, "A huge literature exists for all five theoretical approaches, but the work

necessary to move from research to reality, in terms of the reduction of lower back pain disability, is much slimmer".

Marras *et al.* (1995) have perhaps attempted to address some of these criticisms with a cross-sectional study of manual handling in a wide range of industries. Jobs were analyzed in detail in terms of lifting frequency, horizontal position, vertical position and distance, weight lifted, maximal moments and trunk motions (velocity, acceleration and range of motion in the three cardinal planes). The trunk motions were measured by a Lumbar Motion Monitor attached to the worker's back. The jobs themselves were classified as high, medium or low risk by the number of incidents of low back disorders, which were weighted according to their severity (judged by lost working time). No individual factor was found that had a consistently reliable relationship with high risk jobs, although maximum load moment about L5/S1 was the best indicator. A combination of five factors (lifting frequency, maximum load moment, maximum velocity in the frontal plane, mean velocity in the transverse plane and maximum flexion angle in the sagittal plane) was suggested as being indicative of a high risk of occupationally related low back disorder. A scale was presented, from which the chance of belonging to a job in the high risk category could be determined. It was not made clear exactly why these and not other criteria were included in the combination, especially since the performance of, for example, lifting frequency as an individual measure was not significant. The reason for not selecting some of the variables was explained, but these were often those that performed poorly as individual measures (such as acceleration). Validation was done on 133 jobs, although it was not clear whether

these were from the original 403, in which case they would have contributed to the selection of the five variables. Categorization of the 133 jobs was 78% correct for high risk and 87% correct for low risk, but only 49% for medium risk, which suggests that the model is better at predicting extremes of risk. Further validation is probably required and the authors themselves suggest that longitudinal studies would be helpful.

2.4 BIOMECHANICAL MODELLING AND STRENGTH PREDICTION.

Biomechanical models (Chaffin and Andersson, 1991) can be used to predict the loads and torques on parts of the body, both in situations that have not previously been met (removing the need for extensive data collection) or where direct measurement cannot be achieved. They may also be used to predict the strength of an individual in a particular task. Comparison of a model with the real situation gives greater insights into the functioning of the biomechanical system, particularly if modifications of the model are undertaken to improve its accuracy, allowing us to come closer to a more complete understanding of the real system. It is inevitable, however, that a biomechanical model requires simplifications and assumptions to be made that may affect the validity of the data.

Many types of biomechanical model have been developed. Some focus on a specific part of the body (e.g., the back or lower limb), and some consider the whole body. In general, they consist of two parts (Delleman *et al.* 1992):

- a) A free body diagram. This enables the calculation of the external moments and forces caused by the weight of the body segments and any applied load. Validation is usually done by the comparison of predicted and actual ground reaction forces.
- b) A distribution (internal) model. The external forces and moments must be counterbalanced by internal forces and moments generated by internal structures (e.g., muscles, ligaments, intervertebral discs). Validation is done by comparing predicted muscle action with measured emg activity or predicted spinal compression with that measured *in vivo*.

One of the earliest and best-known models was that of Chaffin and his co-workers, which led to the development of the University of Michigan two- and three-dimensional static strength prediction programs, used widely in the ergonomic community. These types of model presume that the limiting factor in a task is either the capability of a major joint to exert torque, or the ability of the lumbar spine to withstand compressive forces. For any particular task, the computerised model therefore compares the calculated joint torque requirements and spinal compressive forces with previously measured joint maximal torque capability (where no allowance for two-joint muscles was made) and published data on spinal stresses. An adjustment for the effects of intra-abdominal pressure was also included. Early variants (Chaffin, 1969; Chaffin and Baker, 1970) treated the body as a series of six to eight links. Variables entered into the computer included the position and weight of the load, the posture of the subject (using angles between

body segments measured from photographs), the maximum voluntary torques of five major articulations (elbow, shoulder, hip, knee and ankle), and the lengths and weights of the segment links. By incremental loading of the weight at the hand and recalculation of the joint moments with each increment, it was possible to compare the torques calculated for each joint with the maximum measured previously and therefore discover the limiting factor in a particular posture or lift. An analysis of the compression forces on the lower spine was also carried out to compare the maximum forces encountered with data gathered from *in vivo* tests.

Later developments (Martin and Chaffin, 1972; Garg and Chaffin, 1975) incorporated a wider range of body postures and culminated in the two commercially available packages mentioned above. Input required from the user is the anthropometry and posture of the subject and the direction and magnitude of the load. Validation studies of the programs yielded good correlation (r^2 from 0.85 to 0.88) for the 2D model (Chaffin, 1987) but not (r^2 from 0.50 to 0.75) for the 3D program (Chaffin *et al.* 1987). Comparisons of 3D model predicted strength (Chaffin and Erig, 1991) against that measured by Warwick *et al.* (1980), where maximal strengths were measured for subjects exerting forward, backward, to the left and to the right, showed "a consistent 3.7% bias towards over-predicting the mean strengths in these types of exertions". More striking inaccuracies were revealed in a sensitivity analysis where the anthropometry and the postural angles entered into the program were systematically changed. Changing the stature or body weight with a consistent posture tended to result in a prediction of greater strength for persons of smaller stature or lighter body weight. The sensitivity

analysis conducted on the postural angles gave dramatic errors, the worst ($\pm 30\%$) being for forces directed vertically downward when the postural angle of the limiting joint was changed by $\pm 10^\circ$. Marked variation was also noted in vertical lifts and horizontal pulling, but the errors were lower for pushing. It appears that the 3D static prediction program is extremely susceptible to inaccuracies in the measurement of postural angles.

In an attempt to evaluate forces and moments acting on the body during manual handling activities encountered in everyday life, recent biomechanical models have become more complex so that dynamic as well as asymmetric tasks can be assessed. Freivalds *et al.* (1984) designed a two-dimensional dynamic model that takes into account the inertial and accelerative properties of the segments.

Validation of the model was done by correlating predicted ground reaction forces calculated by the model during dynamic lifting tasks with those measured by a force plate ($r = 0.43$, $p < 0.05$). Kromodihardjo and Mital (1986) developed a three-dimensional kinetic ten segment link model. A comparison between predicted and measured ground reaction forces here gave a correlation coefficient of 0.65 ($p < 0.001$). Both of these models were used to analyze lifting tasks with differing task variables such as load size and weight (Freivalds *et al.* 1984; Mital and Kromodihardjo, 1986). Compressive and shear forces on the spine and ground reaction forces during the lift could be plotted against time to evaluate the effect of the variables on these parameters. A similar three-dimensional dynamic model was presented by Chen and Ayoub (1988), but no correlation coefficient was given for the predicted and measured ground reaction forces.

A more extensive validation procedure was attempted for a two-dimensional seven link biomechanical model designed by researchers from the Netherlands (de Looze *et al.* 1992). Not only did they correlate predicted and measured ground reaction forces ($r = 0.88$), but also calculated inter-segmental (joint) reactive forces and moments successively, taking inertial and gravitational effects into account. The analysis was started once at the feet/force plate and once at the hands/load. Once the calculation is complete (i.e., has reached the hands/load or feet/force plate respectively), there should be no "residual" external forces or moments at the hands/load (other than the load itself) or moments at the feet. Because there were only small residual force errors, but the magnitude of residual moment errors was greater and similar in each individual, the resulting conclusion was that the errors were due to inaccuracies in the estimation of the position of the centre of gravity and joint rotation centres. Forces and moments at the L5/S1 joint calculated by starting the analysis at both ends of the linked segment chain were also compared ($r = 0.99$).

This research group compared the same dynamic model with the Michigan 2D static model for a simple dynamic lowering and lifting task (de Looze *et al.* 1994). They found that the peak lumbar moments calculated in each model were comparable when the task was carried out at slow speeds (as suggested by the designers of the static model), but the higher the speed at which the lowering/lifting was carried out, the greater the difference between the two models.

2.4.1 Distribution models.

Distribution (internal muscle) models are included in biomechanical models to consider the counterbalancing internal forces and moments provided by active elements (muscles) and passive structures (e.g., ligaments, intervertebral discs, relaxed stretched muscles). To estimate muscle forces, muscle cross-sectional areas and lengths of moment arm are required. With the availability of modern scanning methods that allow soft tissues to be more clearly visualised, investigators have been able to measure the moment arms of some muscles *in vivo* using computerised tomography (Nemeth and Ohlsen, 1989) or magnetic resonance imaging (Rugg *et al.* 1990). These techniques may prove useful in the future for increasing the accuracy of input anatomical data, since this has been shown to affect predicted compression forces (McGill and Norman, 1987, Bogduk *et al.* 1992).

When many muscles are considered in a distribution model, the contribution of each muscle cannot be directly calculated, because there are too many variables for the number of equations used to describe a static situation - the so-called statically indeterminate problem. Optimization techniques such as linear programming (Seireg and Arvikar, 1973) are utilized, whereby minimization of chosen variables (e.g., muscle tension, moments at the joints (Seireg and Arvikar, 1973) or compressive force on the joint (Schultz and Andersson, 1981) allows a unique solution to be obtained. Linear programming assumes that co-contraction of antagonists does not occur, whereas non-linear programming allows antagonistic

activity to be included (McGill, 1992).

The degree of complexity of distribution models depends on the number of active or passive elements considered. Simple models, such as that by Schultz and Andersson (1981) consider only the internal muscle forces acting across the lumbar spine (Figure 2.3), while ignoring passive elements and assuming no antagonistic muscle action. It has been shown that antagonistic muscle action does occur (Morris *et al.* 1962; Schultz *et al.* 1982b; Schultz *et al.* 1982c; Pope *et al.* 1986; Marras and Mirka, 1988; Lavender *et al.* 1993a; Lavender *et al.* 1993b) and that the amount of force required to move a joint or segment may be under- or over-estimated if the resistance provided by passive structures (Vrahas *et al.* 1990) or intra-abdominal pressure (Morris *et al.* 1961; Davis and Troup, 1964; Davis, 1981) is not taken into account. Greater anatomical accuracy may change spinal compression estimates by up to 35% (McGill and Norman, 1987). Later biomechanical models (Gracovetsky *et al.* 1981; McGill and Norman, 1986; McGill, 1992) have attempted to include some of these variables. For example, the study of McGill (1992) describes a complex model for dynamic, asymmetrical situations, taking into account the co-contraction of agonists and antagonists and the supporting forces and moments provided by passive tissues such as ligaments and intervertebral discs. This model predicted substantially higher compressive loads on the lower back during application of lateral flexion moments than that of Schultz and Andersson (1981).

Figure 2.3: Ten muscle equivalent model for estimating internal force. The ten unknown muscle forces and the intra-abdominal pressure (P) are required for the calculation of the compression (C) and shear (S_a and S_r) on the disc (from Schultz and Andersson, 1981).

The most anatomically detailed model of the lumbar spinal muscles was described by a group from Australia (Bogduk *et al.* 1992; Macintosh *et al.* 1993), who modelled 49 fascicles of the lumbar erector spinae and multifidus, using detailed anatomical data from dissection of cadavers and x-rays of nine subjects. They calculated the extensor moments, compression forces and anteroposterior shear forces due to these muscles in both an upright and fully flexed posture. In the upright position, the thoracic fibres of erector spinae contributed 50% of the total extensor moment about the lower lumbar vertebral joints and up to 86% about the upper joints. Maximal compressive forces on the spine of up to approximately 2800N were calculated. The shear was found to be posteriorly directed in the upper four lumbar intervertebral joints, but anteriorly at L5/S1. The most dramatic change occurring with a flexed posture was the magnitude and direction of these shear forces. In L1/L2, L2/L3 and L5/S1, the direction of shear reversed, and the percentage change in all the lumbar joints ranged from -33% to +185%. The authors emphasize the necessity of using a segmented anatomical model rather than a single muscle equivalent to depict the complex actions of the intrinsic muscles of the back.

Chaffin (1988) suggested that the limitations of many of the models have been that not all variables have been taken into account and that further validation studies are required before we can be sure that models closely resemble the *in vivo* situation. Delleman *et al.* (1992) evaluated and compared simple and complex models as a tool for ergonomists and health professionals. They recommended the simple model described by Schultz and co-workers (Schultz and Andersson, 1981) due to

its reasonable validation results even for asymmetric conditions and postures (Andersson *et al.* 1980; Schultz *et al.* 1982a; Schultz *et al.* 1982b; Schultz *et al.* 1982c; Schultz *et al.* 1983). However, this model does not take into consideration passive structures, which might be expected to play a role, particularly in extreme postures (Floyd and Silver, 1955; Basmajian, 1974).

2.4.2 Strength prediction.

Approaches other than those of the Michigan 2D and 3D programs to the prediction of strength have been tried. Laubach and McConville (1969) attempted to relate a variety of anthropometric measures of 77 male subjects with their strength at different joints. The most reliable predictors of strength were lean body mass and body weight. A similar approach was tried by Yates *et al.* (1980), who measured anthropometric variables and five strength parameters of subjects in an attempt to predict whole body lifting strength. By performing a multivariate analysis using the anthropometric and strength parameters as the independent variables and maximal whole body strength as the dependent variable, different regression equations containing the most important three predictive variables for each condition of whole body lift were obtained. The conditions of lift differed in the distance of the load from the ankle, the height of the load and the gender of subject performing the lift. All the variables measured appeared in at least one of the predictive equations, which emphasized the complexity of attempting to predict maximum strength. The method would be of little practical value, because a



separate equation must be developed for every condition of lift. Only two of the twenty-four regression equations contained the variable body weight.

A method of strength prediction that differs considerably from the 2D and 3D static strength programs is based on the interpolation of strength data from measurements of whole body strength in a wide variety of situations (Sanchez and Grieve, 1992). Two groups of nine subjects were used to provide data on maximal reach and maximal one- and two-handed, symmetrical and asymmetrical lifting strength in many postures. The information was then used in developing equations to predict the maximal lifting strength of an individual of known height and weight where the load was at any position within the reach area tested by the original group. A third group was used to test the predictive equations. The correlation coefficient between predicted and measured strength was 0.99, despite the fact that the test group exerted in positions that differed from the original group. The authors point out that the subjects who tested the predictions were of similar background to those who provided the original strength data and results might change with a different population group. It was also found that if strength was expressed as a fraction of body weight, and height and reach were expressed as a fraction of stature, the predictions were gender-free. Lifting isodynes could then be created for an individual for a particular plane (Figure 2.4). Sanchez and Grieve (1992) correlated applied forces in 156 postures with body weight and found that all but 153 were significantly correlated.

Figure 2.4: Lifting isodynes for a person whose body weight equals that of a 95th percentile male for two-handed exertions. The outermost line represents zero force and each subsequent line an increase in 5kg (from Sanchez and Grieve, 1992).

2.5 LIMITING FACTORS IN THE STRENGTH OF EXERTION.

2.5.1 The upper limb.

The upper limb has been identified by many investigators as a "weak link" in the transmission of forces that are exerted at the hand. In Kroemer's experiments (1974), pushing with a flexed elbow reduced the amount of force applied compared to that with an extended elbow, and pushing with one hand (5th percentile force approximately 250N) was weaker than exerting with two (5th percentile approximately 500N). Eliminating use of the upper limb by pushing with the shoulder increased the magnitude of the forces.

When the Michigan 3D static strength program was used by Chaffin and Erig (1991) to validate the data of Warwick *et al.* (1980), the model predicted that the joints most frequently found as the limiting factor in the exertions were the elbow (43%) and the shoulder (38%). Chaffin *et al.* (1983) used Martin and Chaffin's biomechanical model (1972) to predict the limiting factor in their pushing and pulling experiments. At higher handle heights, this was found to be elbow extension. However, in the experiments of Gaughran and Dempster (1956), the second strongest pulling force was exerted with a flexed elbow.

The subjects of Warwick *et al.* (1980) exerted consistently greater forces with two hands (mean = 199N) than with one (139N), as did the subjects of Chaffin *et al.* (1983) (one-handed/two-handed push: 201/302N, pull: 198/235N), while Fothergill

et al. (1991) often found only small differences between one- and two-handed exertions in many postures, particularly at higher handle heights (see Section 2.2, Figure 2.2, p.34).

It has been shown by many authors that grip strength is dependent on the position of the wrist, but opinions are conflicting on which position affords the greatest grip strength. It has been variously described as 35° of extension (Taylor and Schwarz, 1955), 21° of adduction (Hazelton *et al.* 1975), between 0° and 15° of adduction and between 0° and 15° of extension (Pryce, 1980) and between 0° and 10° of adduction (Drury *et al.* 1985). These angles were measured as deviations from the "neutral" position of the wrist, usually described as alignment of the mid-axis of the forearm with the third metacarpal. It is likely that unfavourable positions of the wrist, particularly abduction and flexion, will decrease the maximum grip strength and therefore affect a person's ability to exert forces.

2.5.2 The back.

Some researchers have identified the back as the limiting factor in force exertion (Poulsen, 1970). In the Michigan static strength programs, the ability of the spine to withstand compressive forces is a constraint additional to those provided by maximal joint torques, which do not include the extension capability of the back. In the regression equations of Yates *et al.* (1980), which relate various parameters to maximal isometric lifting strength, back extension strength appears in ten of the twenty-four equations. The comparison of maximal isometric lifting strength to

back extension strength gave correlation coefficients of 0.69 for men and 0.67 for women. The same correlation by Poulsen (1970) gave values of 0.72 for men and 0.78 for women.

2.5.3 Asymmetry.

Several researchers have shown that exertion in directions with a large left or right directional component or while in asymmetrical postures reduces the force that a person is able to apply. The experiments of Warwick *et al.* (1980) showed that bimanually exerted maximal forces directed to the right or left in asymmetrical postures at knee and shoulder height were 61% of forces directed in the sagittal plane under the same conditions. Pinder *et al.* (1995), in an extensive study on whole body strength in all directions of exertion, showed that forces exerted in directions with a lateral component were generally smaller than those exerted in or close to the sagittal plane.

Many studies have investigated the effect of asymmetrical postures on strength. In the experiments of Warwick *et al.* (1980), mean forces in asymmetrical postures were 81% (torso twisted 90° to the right) and 74% (torso twisted 135°) of those in symmetrical postures under otherwise similar conditions. Garg and Badger (1986) measured the maximum isometric forces for subjects lifting from floor level. The forces were 88% (30° twist), 80% (60° twist) and 70% (90° twist) of those measured for a symmetrical posture. The maximum lifting force of the subjects of

Kumar (1990) in asymmetrical postures (60° twist) was 64% of that in symmetrical postures. Vink *et al.* (1992) showed that both lateral flexion and axial rotation of the torso lead to lower maximal trunk extension strengths.

The figures from the above studies appear to differ considerably, but the measurements were taken under varying conditions of height of hands, direction of exertion, reach distance, posture and degree of torso twist.

For dynamic lifting, the maximum acceptable weight of lift measured by psychophysical methods was reduced by 22% (Garg and Badger, 1986) and 8.4% (Mital and Fard, 1986) for a posture involving a 90° twist as compared to a lift in a sagittally symmetrical posture. Garg and Badger (1986) compared the effect of asymmetry on the strength of both dynamic and static lifts, and gave correction factors which could be applied to existing strength data measured from symmetrical sagittal plane lifting to enable the data to be used for asymmetrical situations. However, these factors are taken from a sample of only thirteen subjects, who were college students rather than industrial workers.

Measured electromyographic activity of the erector spinae and external abdominal oblique muscles, and intra-abdominal pressure, which has been shown to be closely related to spinal stress (Mairiaux *et al.* 1984), were in general shown to be higher in lifts involving twists than those in the sagittal plane (Kumar, 1980).

2.5.4 The hand-handle interface.

Various classifications of prehension have been suggested. Taylor and Schwarz (1955) divided the types of grip into six: cylindrical grasp, spherical grasp, hook or snap, tip (between tips of fingers), palmar (between pads of fingers) and lateral (between thumb and side of index finger). The most widely quoted and simplest classification is that of Napier (1956), who described two main types: the power grip and the precision grip, with an additional hook grip in which the thumb plays little or no part. The power grip clamps the held object between the flexed fingers and the palm with the thumb lying more or less in the plane of the palm, while the precision grip (later called precision "handling" by Landsmeer (1962)) allows the object to be held between the flexor aspects of the fingers and the thumb. Grieve and Pheasant (1982a) presented a classification that takes into account the mechanics of the manual task and is therefore more useful for ergonomic purposes. Both hand posture and function are described depending on the degree of contact between hand and object and the extent to which the hand is used in closed or open chain configuration.

The effect of a poor hand-object coupling was investigated by Drury (1986), who used psychophysical methods to demonstrate that an inferior hand-box interface consistently increased the perceived weight. The influence of the type of handle on strength was shown by Fothergill *et al.* (1992), who tested the horizontal pulling strengths of subjects at two handle heights with four different types of handle. The strength of the pull was reduced by up to 65% by a poor hand-handle interface.

The higher handle height also decreased the strength of the pull, but this effect was reduced when the hand-handle interface was poor (4% decrease) compared to when it was good (45% decrease). The interface was presumably then the limiting factor in the exertion, rather than the height of the handle.

Many attempts have been made to describe an "ideal" handle, but the criteria used for judging do not always produce compatible results. Measurements of grip strength (Ayoub and Lo-Presti, 1971; Pheasant and O'Neill, 1975), pulling strength (Bobbert, 1960; Cochran and Riley, 1986; Fothergill *et al.* 1992), electromyography and fatigue (Ayoub and Lo-Presti, 1971), axial torque (Pheasant and O'Neill, 1975; Cochran and Riley, 1986) and psychophysical factors (Drury, 1980; Drury and Pizatella, 1983) have all been used to determine optimum characteristics of a handle. In a series of papers, Drury and co-workers (Drury, 1980; Drury and Pizatella, 1983; Drury, 1985; Deeb *et al.* 1986) summarized the properties of a good handle:

- a) size: 115mm in length, 25 - 40mm in diameter with a clearance of 30 - 50mm
- b) shape: not as important as other factors
- c) texture: non-slip, but not abrasive
- d) position on a box: asymmetrical, with optimal position for wrist deviation (i.e., 0° - 10° of ulnar deviation).

Cochran and Riley (1986) performed an extensive study of handle shapes and sizes for six different directions of force exertion and found that the optimal shape or size of the handle was dependent on both the task for which it was required and, in some cases, on the gender of the user. They suggest that a specific handle shape and size should be selected if a tool is to be used for specific activities.

2.5.5 Friction.

Frictional considerations may limit a person's ability to exert manual forces. The limiting coefficient of friction (μ) is the ratio of the maximum tangential force (T) that can be exerted on a surface without slipping, to the normal force (N) acting on the surface, or:

$$T = \mu N$$

For a person of low body weight (small N), the value of T on certain surfaces (i.e., with a low μ) need only be small before slip would occur. Grieve (1979b) presents the case for pushing and pulling on sloping surfaces, where vertical and horizontal components of force at the feet contribute components of both normal and tangential force. In these circumstances, the limiting coefficient of friction changes in value, and equations are given for these apparent coefficients. A "slip chart" was also designed (Grieve, 1983b), which allows the minimum possible coefficient of friction for a particular manual task to be read directly from the chart.

Kroemer (1974) gives coefficients of friction for a wide variety of floor surfaces and shoe materials, and Chaffin and Andersson (1991) publish a table from the French National Institute of Safety giving recommendations for combinations of shoe and floor interfaces. Leclercq *et al.* (1995) have shown that extended use of floor surfaces and wearing-in of shoe soles may affect the coefficient of friction, usually increasing its value.

2.6 AXIAL TORQUE.

Many manual handling or sporting tasks involve the body being subjected to twisting forces in the horizontal plane. There is little in the literature explaining how these forces are transmitted through the body, although in recent years some attention has been turned to rotational forces and moments, particularly in the back and the leg.

2.6.1 The back.

Schultz *et al.* (1982a) showed that tasks carried out while subjects were in postures of lateral flexion or twisting of the trunk were not particularly stressful in terms of loads on the spine, particularly when compared to the loads imposed by flexion. However, this is in contrast to a later investigation by the same group (Schultz *et al.* 1983), where resisting twisting moments applied to the upright trunk was the most stressful exertion in terms of spinal compression, muscle contraction intensity and muscle contraction forces. They suggested that resisting large applied forces

isometrically while in an upright posture imposes greater stress on the lumbar structures than the act of rotating while supporting low weight loads.

Pope *et al.* (1986) carried out an electromyographic study to determine the muscles responsible for axial rotation of the trunk. They determined that the internal abdominal oblique muscle ipsilateral, and external oblique contralateral, to the direction of rotation contributed approximately 60% and 20% of the moment of force respectively. The remaining 20% was provided by the ipsilateral rectus abdominis (14%) and erector spinae (6%). Basmajian (1974) mentions similar findings for the oblique abdominal muscles, but found that moderate activity of the erector spinae was present on both sides during axial trunk rotation.

2.6.2 The lower limb.

Although many investigators have measured the range of axial rotation of the tibia on the femur (Hsieh and Walker, 1976; Markolf *et al.* 1976; Mains *et al.* 1977; Markolf *et al.* 1981; Shoemaker and Markolf, 1982; Louie and Mote, 1987; Mills and Hull, 1991), only one study has been found that examines the production of axial torque in the leg (Shoemaker and Markolf, 1982). Maximal internal and external rotation torque was measured (Figure 2.5) with various postural combinations of knee flexion, hip flexion, and internal and external rotation. They found that internal rotation of the leg was generally stronger than external. With the hip, knee and ankle flexed to 10°, 20° and 90° respectively, and the ankle in the neutral position (i.e., neither adducted nor abducted), the mean torques were

Figure 2.5: Mean internal and external rotation isometric torques with the foot in the neutral position. Levels of significance are indicated for differences between paired comparisons of internal and external rotation (from Shoemaker and Markolf, 1982).

31Nm (s.d. 10) for medial rotation and 20Nm (s.d. 7) for lateral rotation. With the foot in a position of maximal external rotation, the medial rotation torque increased to 41Nm, while when the foot was in a position of maximal internal rotation, the lateral rotation torque increased to 29Nm.

2.7 ELECTROMYOGRAPHY.

Although methods for recording the electrical activity of muscles have been available for longer, analysis of specific muscles or muscle groups only began in the 1940's with a myoelectric investigation of muscles of the shoulder girdle by Inman *et al.* (1944). Previous methods of studying muscle function had been restricted to palpation or electrical stimulation of particular muscles, evaluation of the muscular deficiencies of paralysed patients, and the examination of origins and insertions of muscles with application of mechanical principles to deduce their role. Electromyography (emg) enabled researchers to explore the activity of a muscle or several muscles while in action. However, both the gathering and interpretation of emg data are not without problems, as will be discussed.

2.7.1 Types of electrodes.

Three major types of electrode are available (Basmajian, 1974). Surface electrodes are non-invasive and convenient, but have the disadvantage that they cannot be used to investigate deeper muscles. A needle electrode consists of a hypodermic needle containing a wire, which is insulated except at the end. The needle is

injected into the muscle and remains in place, with the wire and the needle forming the two electrodes. Since movement can lead to trauma and sometimes pain, this type of electrode is used more commonly in clinical diagnosis. Fine-wire electrodes are injected into the muscle by means of a needle, but the needle is then removed, leaving the bared tips of the two electrodes and the wires embedded in the muscle. This type of electrode lends itself well to studies of muscle activity, because once injected, it is painless, and deeper muscles can also be investigated.

2.7.2 The relationship between muscle tension and emg.

There is no doubt that relationships exist between the amplitude of an emg trace and the tension developed in a muscle that causes the emg, but the relationship is difficult to quantify and is subject to many influences.

Early attempts to quantify the relationship between emg amplitude and muscle tension were done in the 1950's (Inman *et al.* 1952; Ralston, 1961). Initially, a linear relationship between the two was proposed, but later studies (Zuniga and Simons, 1969; Grieve and Pheasant, 1976; Chaffin *et al.* 1980) showed that the relationship could be better described as curvilinear, the emg amplitude increasing progressively with increasing tension. This was thought to be due to differing recruitment of fast and slow twitch muscle fibres (Chaffin *et al.* 1980). At lower levels of muscle tension, slow twitch fibres are recruited, giving a linear relationship. As the muscle tension increases, fast twitch muscle fibres are recruited which have higher muscle action potentials, giving a second linear

relationship. Chaffin *et al.* (1980) mention another reason for the non-linearity given by Møller (1966) as being the distance between electrodes in testing - the greater the distance between the electrodes, the greater the non-linearity.

2.7.3 The limitations of emg.

2.7.3.1 Posture

Several studies have shown that the amplitude of the emg signal depends on posture (i.e., the length of the muscle) (Inman *et al.* 1952; Ralston, 1961; Grieve and Pheasant, 1976; Gerdle *et al.* 1988; Mouton *et al.* 1991). Electrical activity of a shortened muscle may be increased despite relatively low levels of loading, while that of a lengthened muscle may be moderate despite substantial developed torque. A suggested explanation for this is that the duration of the active state may increase in lengthened muscles and therefore fewer action potentials are required for the same degree of mechanical activation (Grieve and Pheasant, 1976).

2.7.3.2 Electrode position

The position of surface electrodes on the skin has also been shown to affect the amplitude of the emg trace. The further away the electrodes are from the centre of the muscle belly, whether longitudinally or transversely, the greater the reduction in the electrode potentials (Zuniga *et al.* 1970).

2.7.3.3 *Cross talk*

Since surface electrodes are placed on the surface of the skin, emg signals recorded from one muscle may be subject to interference from active muscles nearby (Koh and Grabiner, 1991a). A double differentiation technique (Broman *et al.* 1985), which passes the signal through two sets of amplifiers, reduces cross talk significantly (De Luca and Merletti, 1988; Koh and Grabiner, 1991b). Use of this relatively new method may increase the reliability of surface emg values.

2.7.3.4 *Deformation and displacement of fine wire electrodes.*

A possible problem with fine wire electrodes is distortion while they are embedded in muscle. Jonsson and his co-workers (Jonsson and Bagge, 1968; Jonsson and Reichmann, 1969) assessed the displacement (by measuring the length of wire remaining outside the skin), deformation (either once the electrode had been removed, or with radiographs while still *in situ*) and the frequency of fracture of fine wire electrodes. Displacement of inserted wires was up to 17mm (Jonsson and Reichmann, 1969) and the wires were always deformed. However, the distance between the uninsulated part of the electrodes showed only small displacements in relation to each other. These experiments did not attempt to relate the difference in the condition of the electrodes with measured emg values, so the effect of the changes is not known.

2.7.3.5 *Repeatability*

Komi and Buskirk (1970) found that the test-retest reliability on both the same and separate days for measuring emg signals of isometric contractions of biceps brachii was better with surface than with fine wire electrodes. However, a consistent position of the surface electrodes was assured by marking the skin. Grieve and Cavanagh (1974) used various objective methods of recording emg from the lower limb during two consecutive cycles of walking to show that even this repetitive activity was difficult to reproduce in quantitative terms. They suggest that signals should only be compared on a 5 to 6 point scale.

2.7.3.6 *Fatigue*

The effect of fatigue on the emg signal is to increase the amplitude of the integrated emg as more motor units are recruited, to cause synchronisation of action potentials, leading to tremor, and to decrease the frequency (Lippold *et al.* 1960; Basmajian, 1974; Grieve and Pheasant, 1982a). Interruption of the blood supply to the fatigued muscle prevents recovery of the integrated emg signal (Lippold *et al.* 1960). Fatigue in one muscle may cause redistribution of activity to other synergistic muscles (Lippold *et al.* 1960). Fatiguing of subjects during experimental procedures may therefore confound the later analysis of recordings.

2.7.3.7 *Temperature*

Bell (1993) has shown that the surface emg/force relationship may vary with changes in temperature. He showed that exposure to a hot (40°C) environment decreased the emg signal, while it was increased after prolonged exposure to cool (10°C) conditions. Suggested reasons for the changes were fluid shifts within the muscle, a change in conduction velocity and sweating. Similar results were shown by Grieve and Pheasant (1982a), who tested emg signals from the thenar eminence while the gloved hand was immersed in hot or cold water.

CHAPTER 3.

RELATIONSHIPS BETWEEN FORCE AND POSTURE FOR ONE-HANDED EXERTIONS IN MANY DIRECTIONS.

3.1 INTRODUCTION.

When a standing person engages in a static manual exertion, the force and its direction cannot be predicted from the posture because they are a matter of choice. However, a glance at the posture usually conveys a strong impression of both. The reason for this is that there are functional implications of a particular combination of body weight distribution, use of the foot base and hand location. Some of the implications for bimanual exertions in the fore-aft plane were studied in a series of papers by Grieve and Pheasant (Grieve and Pheasant, 1981; 1982b; Pheasant and Grieve, 1981; Pheasant *et al.* 1982). They introduced the so-called Postural Stability Diagram in order to discuss their results (see section 2.1.3, p.25). The Postural Stability Diagram (Grieve, 1979a; 1979b) is a graphical method of representing static horizontal and vertical forces during two-handed exertion. The name emphasizes the equilibrium that exists during a static manual exertion and its link with the chosen posture. It enables the user to consider the equilibrium at the feet (where frictional limitations might permit slip) and at the hands (where the task may impose particular demands) simultaneously.

Figure 3.1a represents the projection in a vertical plane of a person of body weight, W , who is exerting a static manual force, F , together with a torque, T , in the plane of analysis. The locations of the centre of foot pressure (CFP), the body's centre of gravity (CG) and the centroid of force exertion at the hand-handle interface (CH) are shown. The force, F , has been resolved into an upward (UP) and a forward (FORWARD) component. CH is at height, h , above the ground and at a

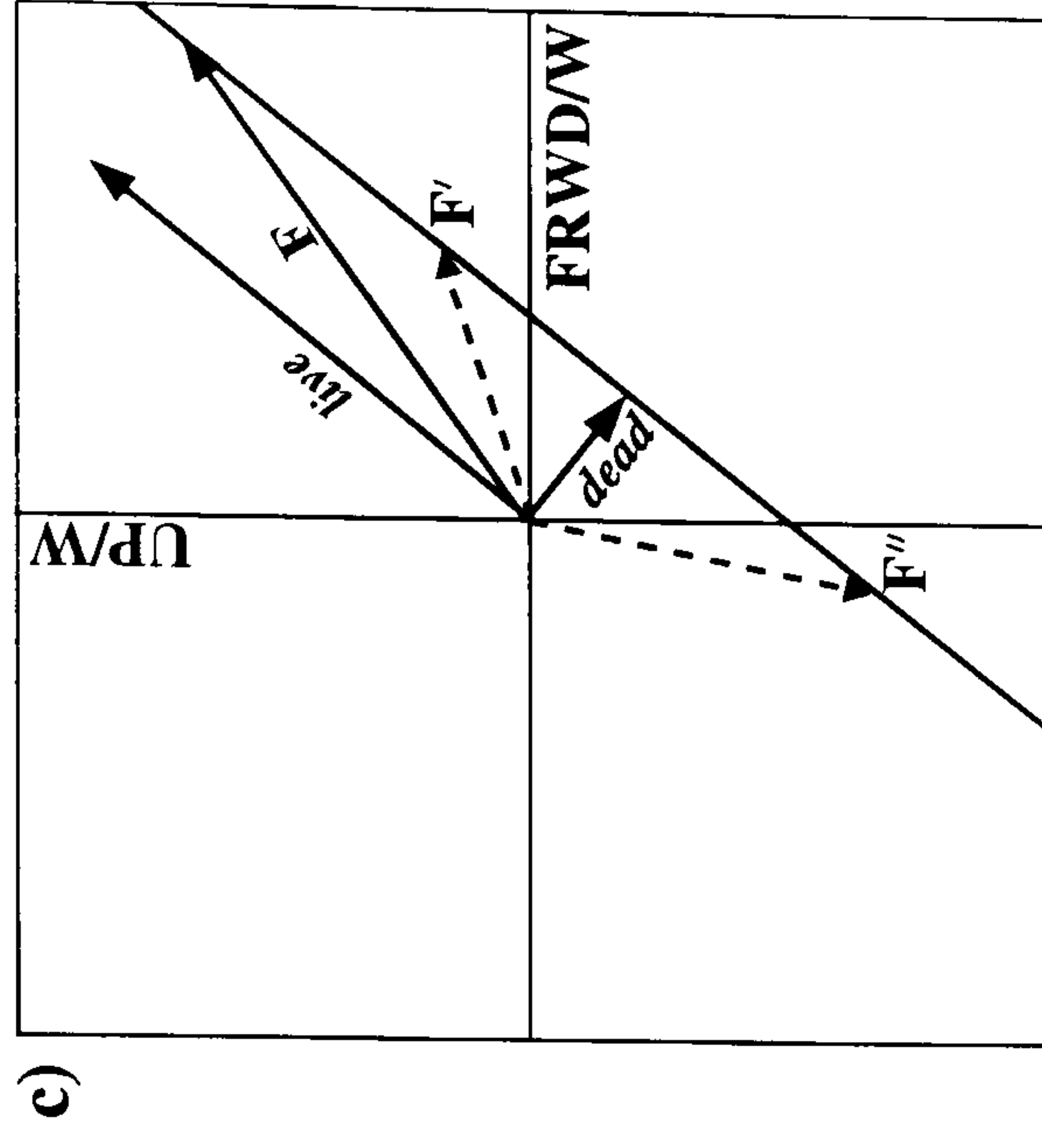
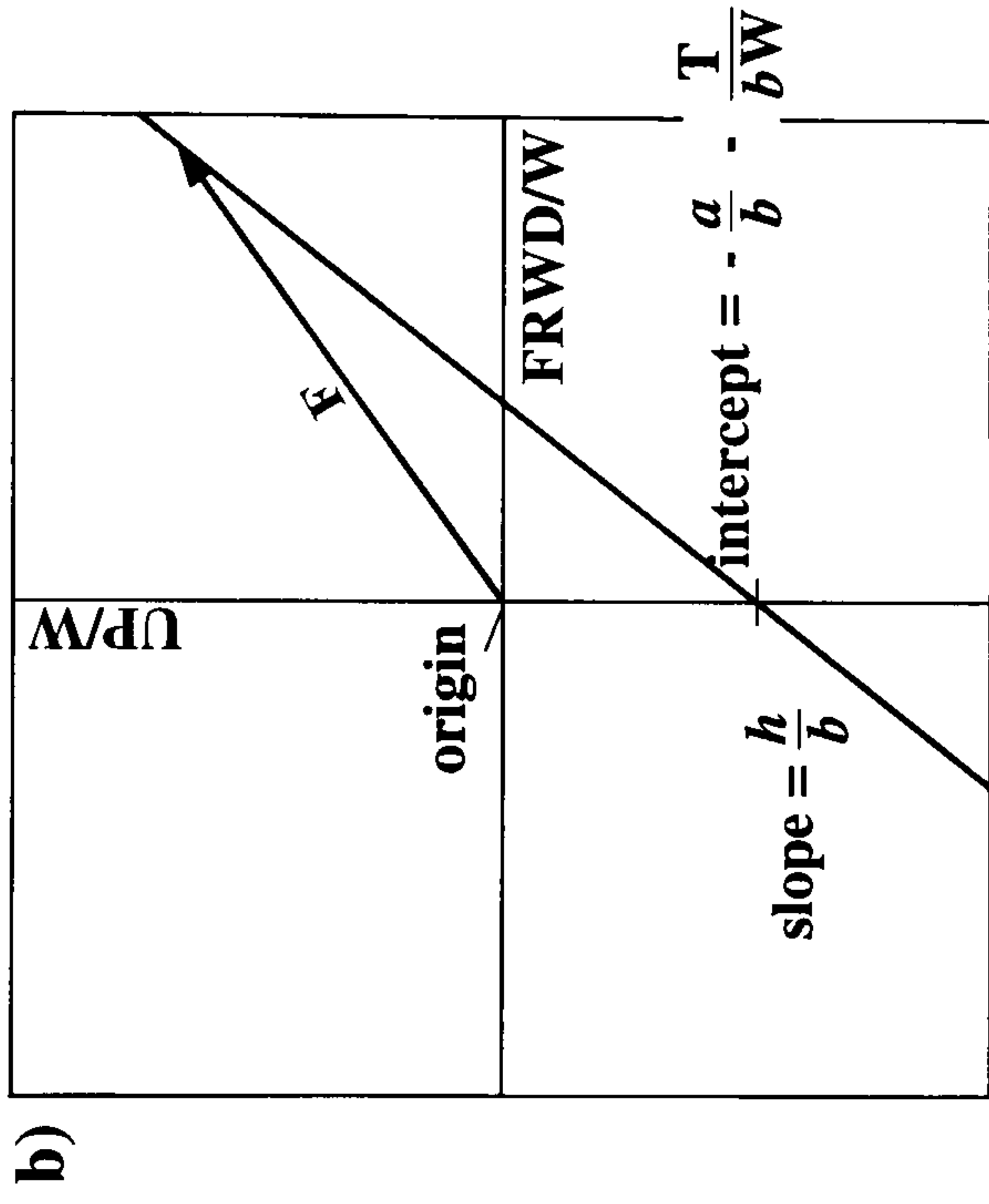
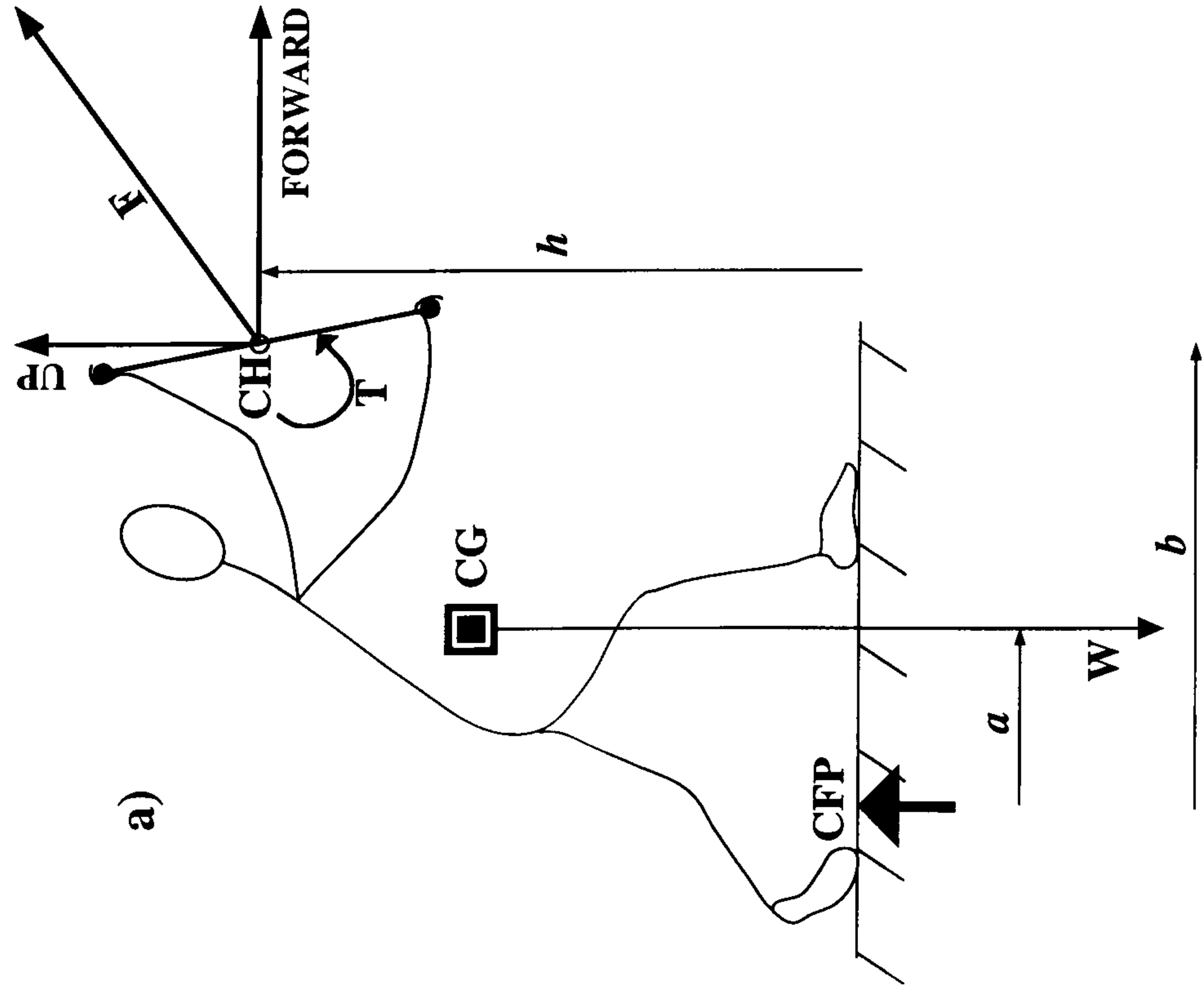


Figure 3.1: Postural Stability Diagram: a) Person exerting up and forward (W - body weight, F - force at hands, T - torque at hands, CG - centre of gravity of body, CH - centroid of force at hands, CFP - centre of foot pressure), b) Postural Stability Diagram with line of equation of static exertion, ($FRWD$ - FORWARD) pressure), c) *live* and *dead* components of force (F' and F'' - possible alternative force vectors to F).

horizontal distance, b , from the centre of foot pressure. The CG is at a horizontal distance, a , from the centre of foot pressure. If the moments due to forces acting **upon** the body are taken around the centre of foot pressure as fulcrum and the reaction to applied manual torque is included, the equation of static exertion is obtained:

$$\frac{UP}{W} = \frac{h}{b} \cdot \frac{FORWARD}{W} - \frac{a}{b} - \frac{T}{bW} \quad (i)$$

The equation is that of a straight line on the Postural Stability Diagram (Figure 3.1b). Its slope (h/b) is the same as that of the line joining CH to the centre of foot pressure and it has an intercept that depends on a) the horizontal dispositions of CH and CG relative to the centre of foot pressure and b) the hand torque, although the latter factor is zero in the current experiments.

It is important to note that the adopted posture and use of the foot base does not determine what force is exerted (which is a matter of choice by the subject), but it does determine what combinations of UP and FORWARD are possible in the circumstances, since any applied force must obey the equation. In order to exert any force of chosen magnitude and direction, the head of the vector must lie somewhere on the line representing the equation of static exertion. In the absence of torque at the hands, moving the body (and therefore the centre of gravity) backward or forward with fixed foot and hand position will change the intercept of the equation, whereas moving the foot and/or hand position changes the slope. Thus many lines (of various slopes and intercepts) could pass through the head of the chosen vector, especially if the force was modest. For any given position of

the hands and feet, lines with intercepts drawn at their extremes on the Postural Stability Diagram show the limits due to posture of the available combinations of UP and FORWARD.

The authors (Grieve and Pheasant, 1982b) also suggested an alternative method of considering the applied force (see section 2.1.3, p.28) by resolving it, not into vertical or horizontal components, but into a component (*live*) along the centre of foot pressure-to-CH axis and another (*dead*) perpendicular to the *live* component (Fig 3.1c). The magnitude of the *dead* component is solely attributable to the deployment of body weight in that posture. The *live* component can be likened to a volitional component chosen by the subject within his or her muscular capacity. It passes through the centre of foot pressure and therefore has no moment about it. The muscle activity providing the *live* force cannot be separated, because maintenance of a posture will also require some muscle action. By altering the *live* component when the body has a particular disposition of weight i.e., a fixed *dead* component, forces of different magnitude and direction (e.g., F' or F'') may be achieved. In every case, the head of the vector lies on the line representing the equation of static exertion.

Postural studies of bimanual symmetrical exertions have been made by several authors (see section 2.1, p.22 above), as well as some involving asymmetry (see section 2.2, p.31). In the current experiments, use of a special handle to measure force and direction without the possibility of torque avoided the complexity that torque introduces into the equation of static exertion.

3.1.1 Aims.

The purpose of this study was:

- a) to provide an insight into postures adopted during one-handed, often extremely asymmetric, exertions in many directions between up and down, forward and backward and left and right;
- b) to investigate the relationship between posture and strength by simultaneous recording of the applied force and its direction together with the posture;
- c) to examine differences in demands made upon the body in asymmetrical as opposed to symmetrical exertion by detailed analysis in both vertical and horizontal planes;
- d) to determine whether the idea of the Postural Stability Diagram can be expanded to incorporate force directions other than those in the fore-aft plane.

3.2 MATERIALS AND METHODS.

3.2.1 Experimental equipment.

At one end of a 20m by 7m room, a rigid scaffold framework supported the right-handed handle of the Tri-Axial Force Measurement System (TAFS) described by Pinder *et al.* (1993), and a platform of dimensions 1.83m by 0.61m, which was covered with emery cloth to prevent slipping. This arrangement (Figure 3.2) enabled a subject to stand on the platform and exert forces on the TAFS handle.

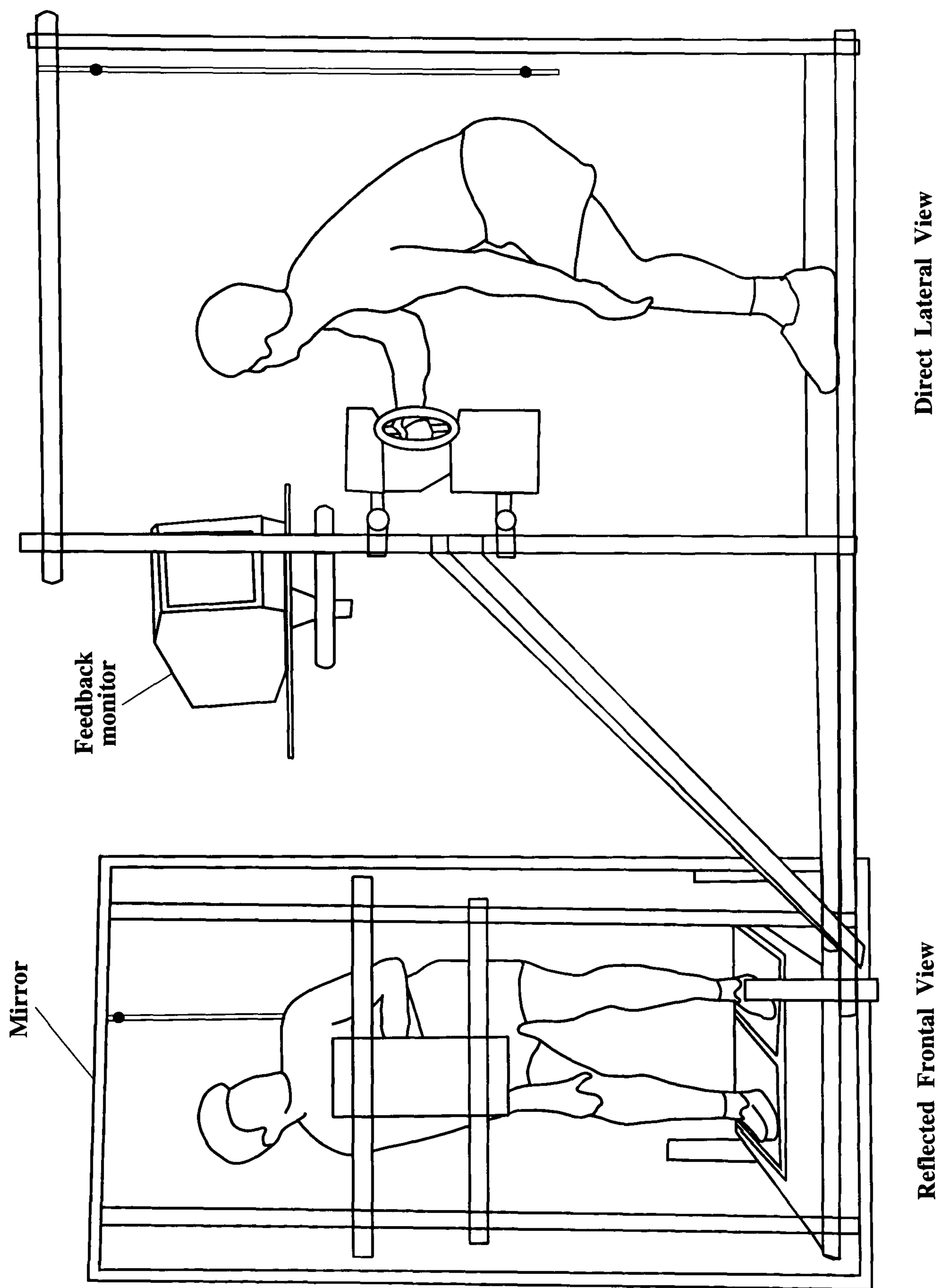


Figure 3.2: Experimental set-up.

The platform was marked with three parallel, longitudinal white lines: one central, and two others 0.26m on either side of it.

The TAFS (Figure 3.3) measures the magnitude and direction of an imposed force while precluding the exertion of a torque in any plane. The handle is supported on a mounting by three rigid links, each containing a ring transducer, on which were fixed foil strain gauges. These convert the strain caused by application of force on the handle to voltages, which were then amplified, digitized by a CED1401 analogue-to-digital converter and passed to a BBC host computer. The TAFS handle was mounted on a framework that allowed adjustment of its height. Output from the BBC computer was displayed on a monitor positioned in clear view of the subject.

A mirror, of height 1.9m and width 1.3m, was placed vertically in front of the subject and angled at 45° to the central longitudinal line of the platform. The centre of the mirror was at a distance of 1.38m from the handle along this line. A plumb-line, consisting of a rod carrying two marker balls (20mm diameter at 1m spacing), was suspended vertically over the central line of the platform behind the subject.

3.2.1.1 *Photography.*

Photographs were taken with a camera, whose optical axis was 1.5m above the ground at right-angles to the platform's longitudinal axis. The optical axis of the

Figure 3.3: The TAFS handle (from Pinder *et al.* 1993).

camera was measured in the following way:

- a) The camera was placed at the end of the room away from the apparatus and its height was adjusted until the middle of the lens was at a height of 1.5m and pointed towards the scaffold framework.
- b) Two plumb-lines, approximately 2m apart, were suspended from the ceiling in a line at right-angles to the longitudinal axis of the platform. One white plastic sphere of diameter 20mm was attached to each plumb-line at a height of 1.5m.
- d) The camera view was then adjusted until it centred on the two spheres with the one further from the camera obscured by the closer. This gave an approximation of the optical axis of the camera, subject to internal camera error.
- e) A small mirror of diameter 60mm was then taped tightly to the front of the camera lens. By standing near the scaffold framework, an observer could see the two spheres and their reflections in the mirror attached to the camera. The position of the camera was adjusted until all four spheres were aligned, so that only the one closest to the observer was visible. The position of the camera was almost identical to that previously, which indicated minimal internal camera error.

The camera was a Canon AE-1 with automatic exposure, focal length set to 150mm and shutter speed 1/60s. Black and white Ilford FP4 film was used. The camera was mounted on a tripod 16.64m from the central line of the platform.

This afforded both a mirror-reflected frontal view and a direct lateral view of the subject (Figure 3.2). Paper backdrops were mounted lateral and posterior to the subject and the workspace was floodlit.

3.2.2 Subjects.

Eleven young men, nine military personnel and two civilians, took part in the study. Their mean age, height and weight were 27yrs (s.d. 3.3), 1.75m (s.d. 0.06) and 75.0kg (s.d. 9.6) respectively. Two subjects were left-handed.

3.2.3 Experimental procedure.

The postural investigation was done together with other experiments, which determined the omni-directional strength of the same subjects (Pinder *et al.* 1995). The study was divided into three five-day sessions, with the handle at 1.0m, 0.5m and 1.5m height respectively. The first four days of each session were spent in gathering strength data, so that by the fifth day, when the postures were photographed, the subjects were thoroughly familiar with the apparatus.

Before the beginning of the photographic part of the session, the procedure was explained to the subjects, who signed a consent form. Subjects wore shorts and trainers.

Each subject stood on the platform with both feet facing forward (Figure 3.2), one

on each side of and equidistant from the central line, which passed below the centre of the handle. The feet did not have to be directly opposite each other, but were not allowed to pass in front of a line drawn on the platform at right-angles to the central line directly beneath the handle centre. Each subject grasped the handle with his right hand.

Postures were recorded during exertions in twenty-six directions. These included vertically up and down, horizontally forward, backward, left and right and twenty directions in between these. They are depicted in a "three-dimensional" representation in Figure 3.4. However, the direction in which a subject was required to exert was represented on the monitor screen with a two-dimensional display. A specified direction was indicated by a small square within a larger rectangle (Figure 3.5). The subject was limited to a force within 5° in any direction from the required direction of exertion. The directions were presented in random order.

The first target square, representing one of the directions, appeared on the monitor screen and the subject would attempt to exert in that direction. Because of 40 minutes previous experience using the handle at that particular height in the preceding four days, the subjects were familiar with relating the direction of exertion to the monitor display. A pattern on the screen, indicating the direction and magnitude of the force exerted, provided feedback. When the subject achieved the required direction, the computer gave an audible signal and a photograph was taken while the exertion was maintained. The force achieved and the exact

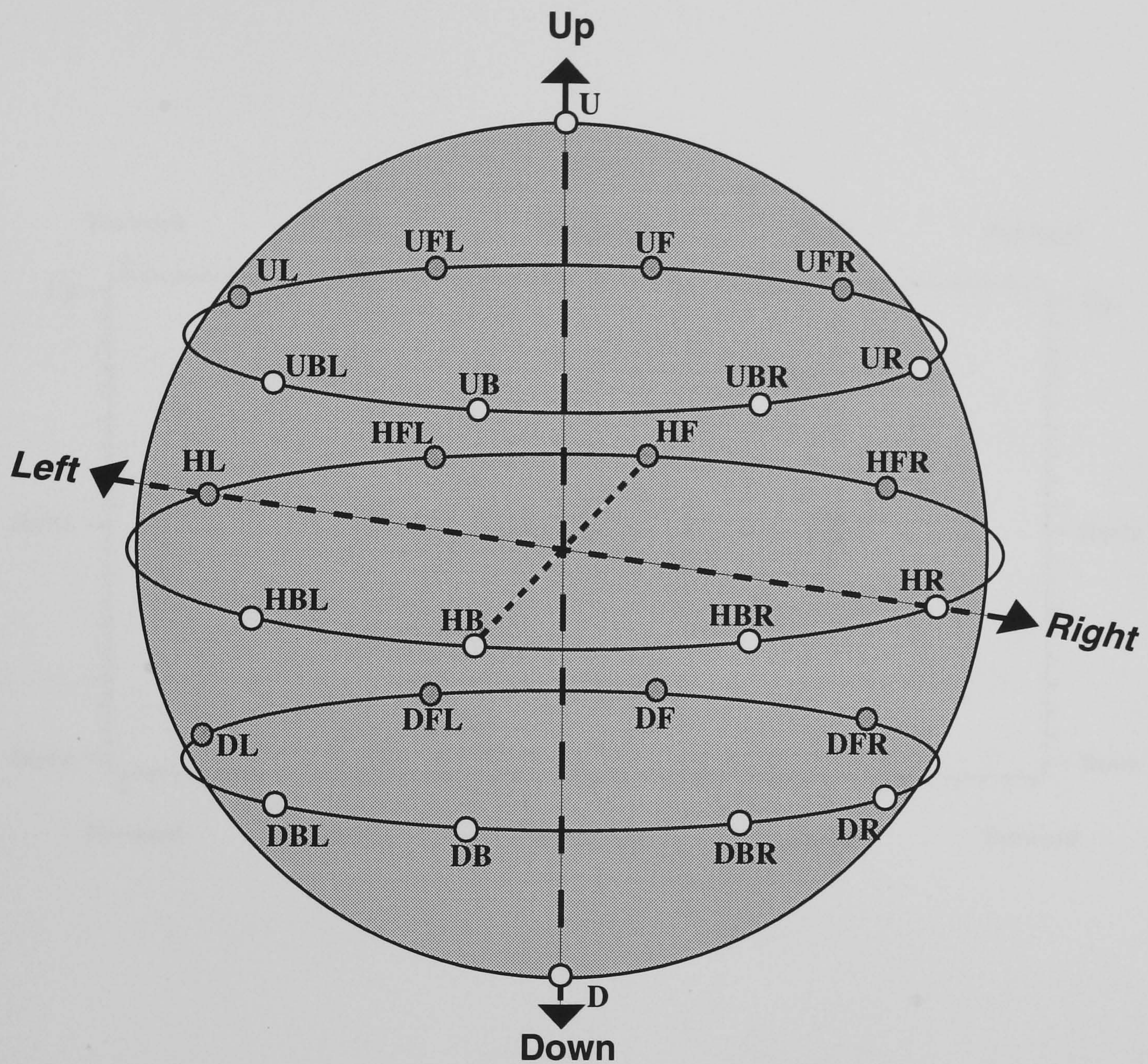


Figure 3.4: The twenty-six directions of exertion. U = up, D = down, L = left, R = right, F = forward, B = backward, H = horizontal.

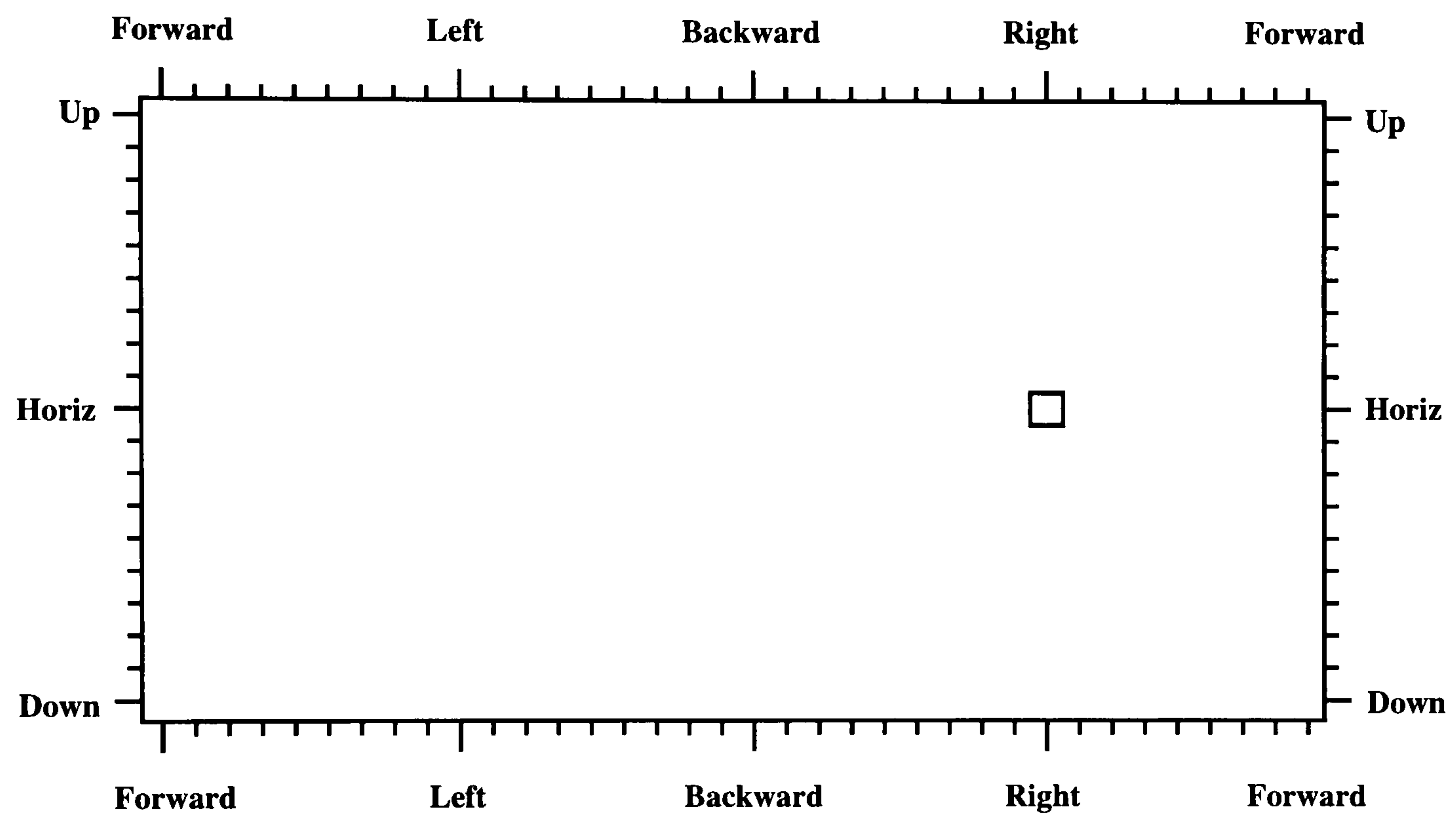


Figure 3.5: An example of the monitor display shown to each subject, including a target set to one of the twenty-six directions (horizontally and to the right).

direction of exertion were recorded by the computer. The force was usually less than the maximum that had been observed in the previous four days. If the subject was unsuccessful after two minutes, a rest was given. If the subject remained unsuccessful after four two-minute periods, that direction was omitted. This occurred on only 9 of the 858 occasions.

3.2.3.1 *Digitization.*

The photographic negatives were mounted in a Leitz 250 Projector, projected orthogonally on to a screen 460mm x 660mm and digitized using a GP-7 GrafBar Mark II sonic digitizer interfaced with a BBC Master computer. Twenty-two points per negative (Figure 3.6), plus the two markers on the plumb-line, were digitized. If the landmark was not clearly visible, an estimate of its position would be made. The location of the handle centre was obscured in the mirror view by its mounting, so a template with the centre marked was used during digitization.

The digitized points for each subject, handle height and direction were rotated and scaled, using the plumb-line markers, to provide horizontal and vertical millimetric co-ordinates. The data were transferred to an Archimedes A5000 computer. A computer program was written to compute the three-dimensional co-ordinates of the landmarks with the handle centre as origin from this information. In order to scale the image of the subject in the mirror so that it was the same size as that in the direct lateral view, each co-ordinate in the mirror view was multiplied by a

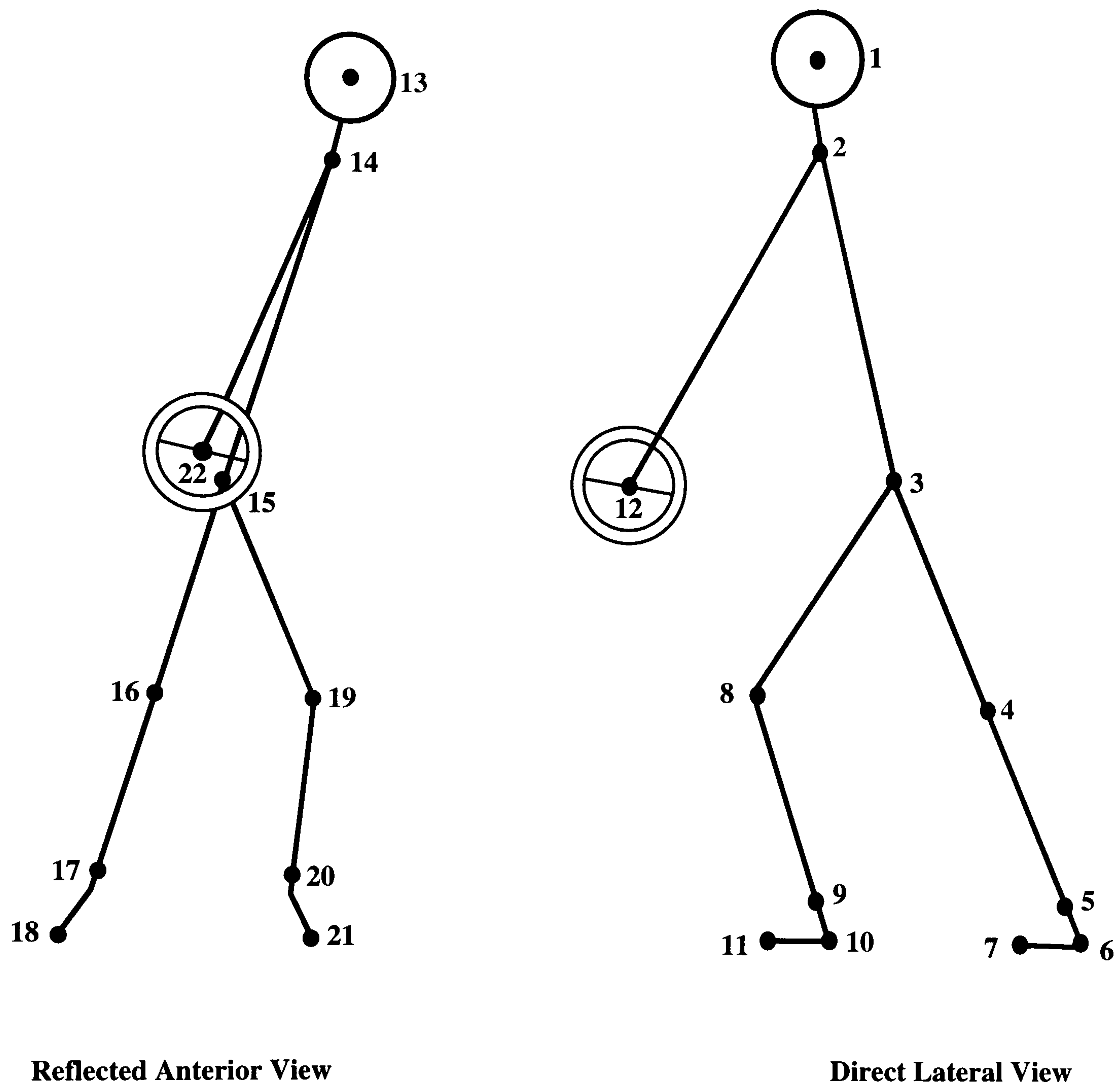


Figure 3.6: Digitized points: 1,13 - head; 2,14 - C7; 3,15 - mid-hip; 4,16 - left knee; 5,17 - left ankle; 6 - back of left foot; 7 - front of left foot; 8,19 - right knee; 9,20 - right ankle; 10 - back of right foot; 11 - front of right foot; 18 - lateral side of left foot; 21 - lateral side of right foot, 12,22 - handle centre.

factor (distance from camera of co-ordinate in virtual image / distance of central line of platform from camera).

The exact direction and the magnitude (as a percentage body weight) of the applied force was obtained from the TAFS handle, and the CG of the body was calculated from anthropometric tables (Pheasant, 1986, modified from Dempster, 1955). With this information, it was possible to determine the position of the centre of foot pressure by using equations of static exertion for three dimensions. These were obtained by taking moments in three planes about the centre of foot pressure (CFP) of the forces acting upon a hypothetical person exerting upward, forward and to the right on a handle (CH) at height h (Figure 3.7). In the vertical left-right plane, the centre of foot pressure is at a horizontal distance, c , from the CG of the body and the hand centroid is at a horizontal distance, d , from the centre of foot pressure. In the vertical fore-aft plane, the centre of foot pressure is at a horizontal distance, a , from the CG of the body and the hand centroid is at a horizontal distance, b , from the centre of foot pressure (as in Figure 3.1a). The components of the force exerted by the subject are given by UP, RIGHT and FORWARD. An additional factor is the horizontal moment at the feet, given by M. The equations are:

$$\frac{UP}{W} = \frac{h}{b} \cdot \frac{FORWARD}{W} - \frac{a}{b} \quad (ii)$$

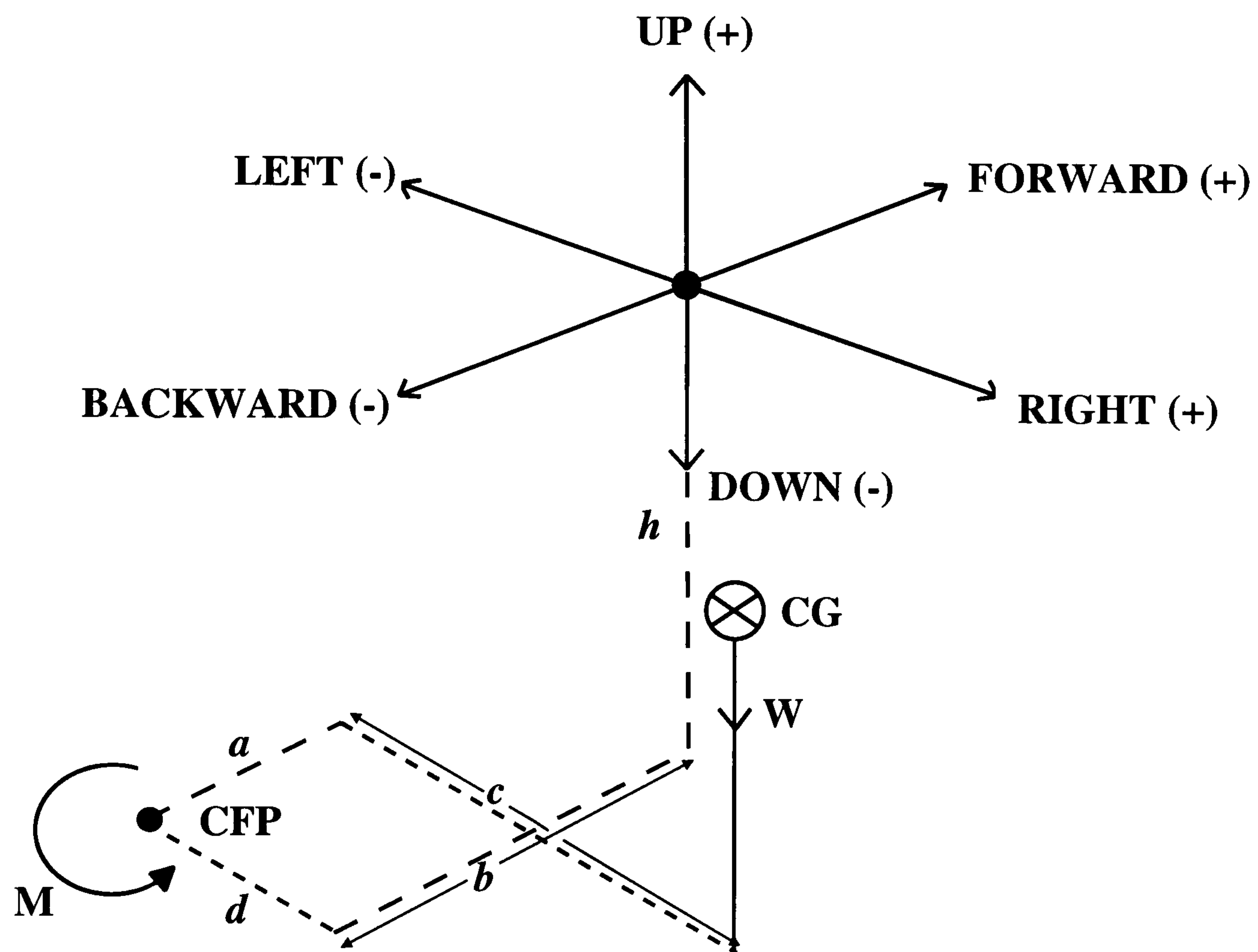


Figure 3.7: Conventions used for three-dimensional exertion. CG = centre of gravity, CFP = centre of foot pressure, M = horizontal moment at feet, W = body weight.

$$\frac{UP}{W} = \frac{h}{d} \cdot \frac{RIGHT}{W} - \frac{c}{d} \quad (iii)$$

$$\frac{FORWARD}{W} = \frac{b}{d} \cdot \frac{RIGHT}{W} - \frac{M}{dW} \quad (iv)$$

Since a limited number of landmarks were digitized, assumptions were made when calculating the CG of the upper limbs. The right hand was known to be in contact with the handle, so the position of the CG of the right upper limb was calculated as being halfway between C7 and the handle in both the y (fore-aft) and z (up-down) axes and halfway between the right knee and the handle in the x (left-right) axis (Figure 3.8). The left upper limb was assumed to be hanging by the left side, so the position of the CG was calculated as being the same as the left knee in the x axis, C7 in the y axis, and halfway between C7 and the handle in the z axis.

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The errors associated with the above assumptions were considered by postulating more extreme postures as follows: the left upper limb was assumed to have 90° shoulder abduction in the left-right plane and a fully extended elbow. Its CG thereby preserved the same y and z co-ordinates as C7. The right upper limb CG was assumed to have the same y and z co-ordinates as before, but its x co-ordinate was assigned to that of the right knee. The effect of these extreme assumptions (for all subjects, positions and handle heights) on the calculated positions of the CG and centre of foot pressure was a mean difference of 39mm (s.d. 7) and 38mm (s.d. 10), which suggests that the error associated with the more modest

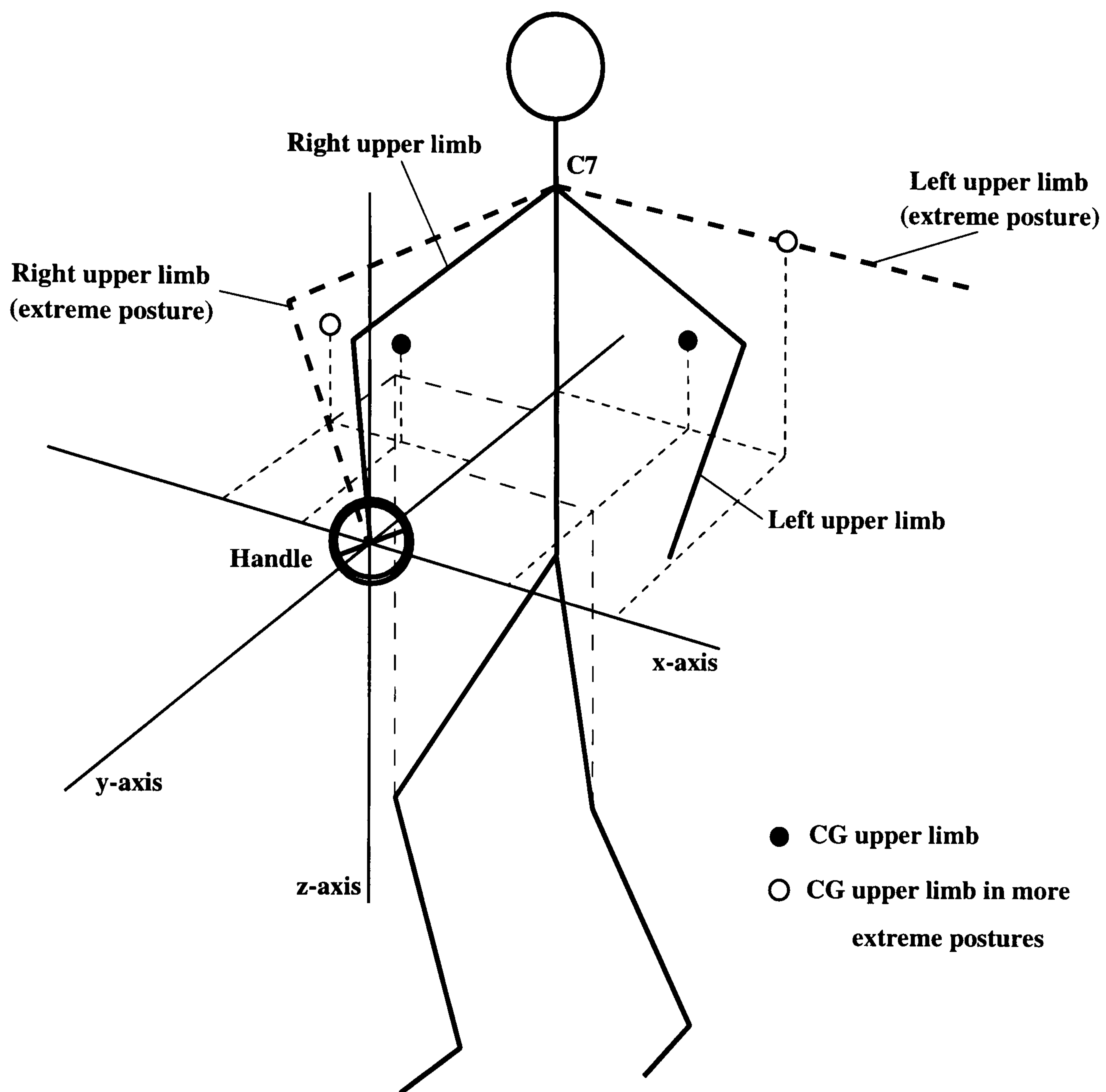


Figure 3.8: Positions of CG of upper limbs used in calculations compared to those in a more extreme posture used to assess the magnitudes of possible errors.

assumptions was acceptable.

For postural analysis, the co-ordinates were projected into the vertical plane containing the force vector so that two-dimensional representations for each handle height, force direction and subject could be constructed. For example, if the exerted force was forward and upward (or any other direction in the fore-aft plane), the resulting projection of the posture was in the fore-aft plane (as in Figure 3.1a). A posterior projection was obtained from a subject exerting horizontally and to the right (or any other direction in the left-right plane).

3.3 RESULTS.

3.3.1 Posture.

The forces exerted by the subjects varied considerably. The mean forces achieved (Figure 3.9) were greater than had been observed in the previous four days on only 8 out of 78 occasions (26 directions at 3 handle heights). The strength measurement protocol (Fothergill *et al.* 1993) had given a total period of 40 minutes for maximum forces to be reached in all directions. It had been found that the increase in mean force over the entire range of directions after that time was less than 2.6%. Occasionally, the force exerted in one particular direction may not have reached maximum, but if asked to exert specifically in that direction (as with the postural study), a person might then be able to apply a greater force.

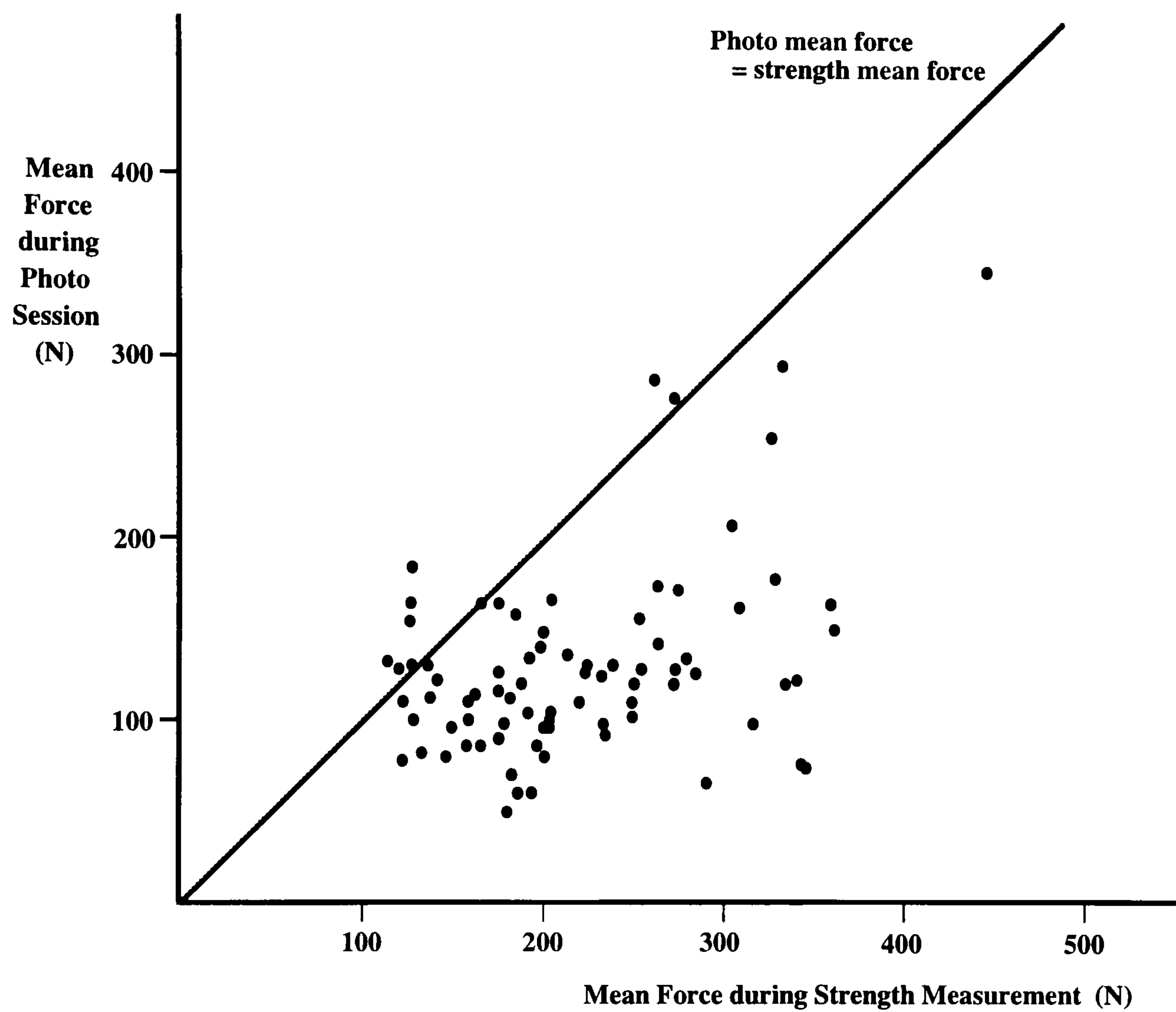


Figure 3.9: Mean force of the eleven subjects during strength measurement vs mean during photo session (26 directions at 3 handle heights).

It was not feasible to ask the subject to achieve both the desired direction and the maximum (i.e., previously achieved) force; to achieve both in every direction would have been logistically unacceptable and fatiguing. The advantage of analyzing the posture that existed when only a chosen direction was achieved is that examples of both weak and strong exertions were found and that differences associated with the magnitude of exertions could be studied.

Figure 3.10 shows stick figure representations of the eleven subjects, viewed in the plane of exertion (i.e., posteriorly), exerting horizontally and to the right at all three heights. The stick figures are arranged from the weakest exertion on the left to the strongest on the right. The calculated CG and centre of foot pressure of the subjects are also marked on the diagrams, showing the progressive increase in the horizontal distance between the two as the exerted force increases.

A change from the postures of those with the weakest to those with the strongest forces can be seen in all the subsets of the figures, but a pattern is particularly noticeable where the subjects exerted horizontally and to the right at a height of 1.0m. As the force becomes greater, the subjects are leaning further away from the handle, with the centre of foot pressure close to the left foot and the CG well to the right. By using the body weight in this way, the force component would consist of a greater proportion of *dead* force. The greatest observed exertion at 1.0m height was 43% of body weight. At 1.5m, subjects had less opportunity to deploy body weight and a greater proportion of the force would have been *live* force. Even though the centre of foot pressure was usually close to the left foot,

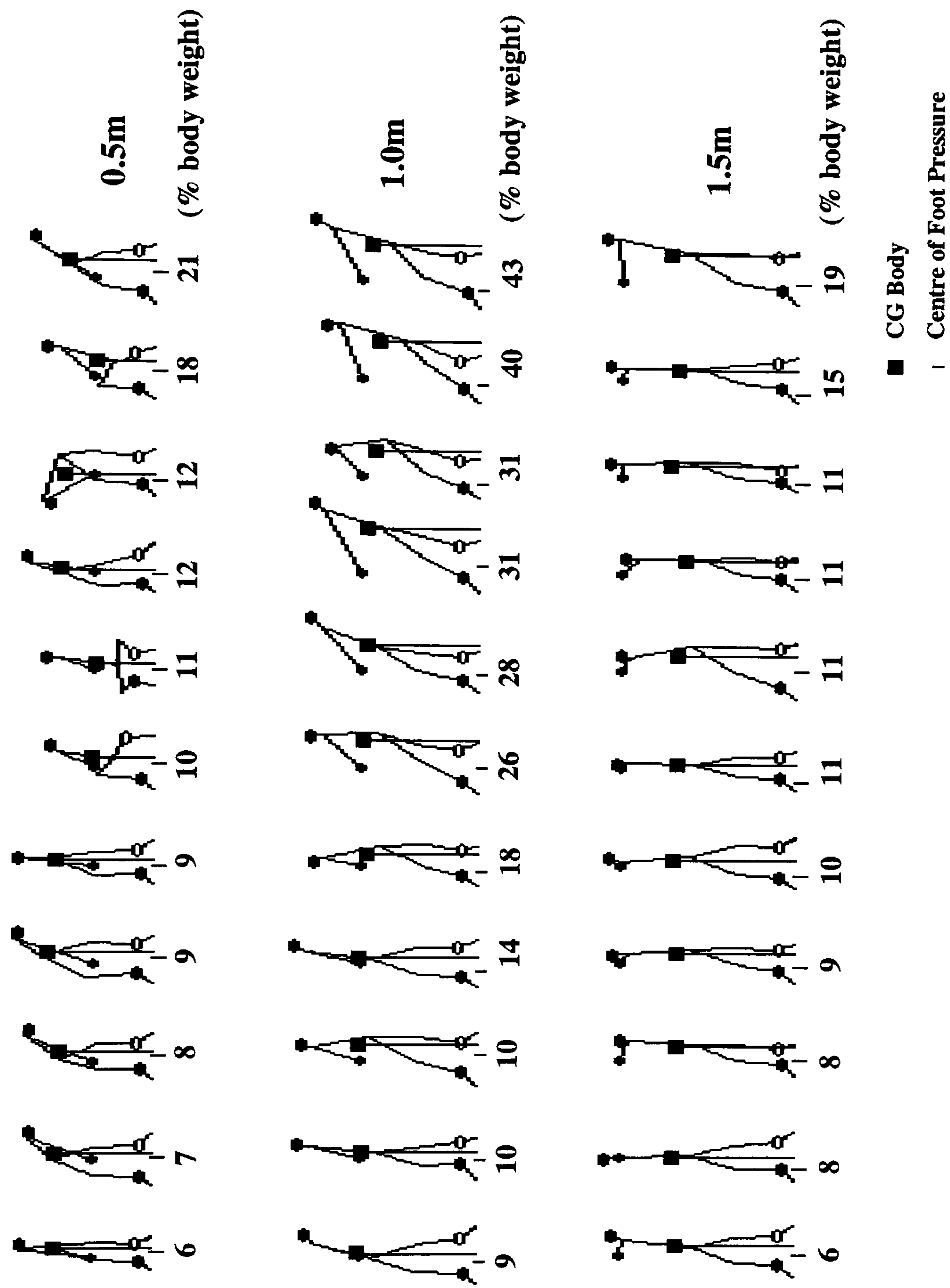


Figure 3.10: Postures of all subjects exerting horizontally and to the right at the three handle heights, showing forces achieved as a percentage of body weight.

the maximum observed force was only 19% body weight.

Figure 3.11 is a similar collection of postures, but the direction of exertion is now downward, backward and to the left, a difficult direction to attain with the right hand. Here, the projection of the postures is oblique postero-lateral (an open circle at the ankle represents the right lower limb, while a closed circle represents the left). Again, a difference can be noticed between subjects who exerted greater and lesser forces. With the handle height at 1.0m, the subject with the smallest force (6% body weight) stood straight-legged and upright. The subject with the greatest force flexed both knees to an acute angle and leaned backward, so that his centre of gravity was low and even his head was below handle height. This suggests optimal body weight deployment and the subject achieved a force of over half his body weight. A similar principle was applied by the two subjects with the greatest force at handle height 0.5m.

Figure 3.12 shows the set of postures of the subjects exerting upward and forward. The subjects exerting the greatest force at handle height 0.5m are in a crouching posture, pushing upward and forward from below. At a handle height of 1.0m, there is an obvious pattern from the weakest to the strongest. The subjects with the strongest exertion are attempting to align their bodies with the direction of the force vector and the force is largely *live* component.

Figures 3.13 to 3.16 show more examples of posture sets. The entire collection of stick figures depicting the postures of subjects at all handle heights and in all

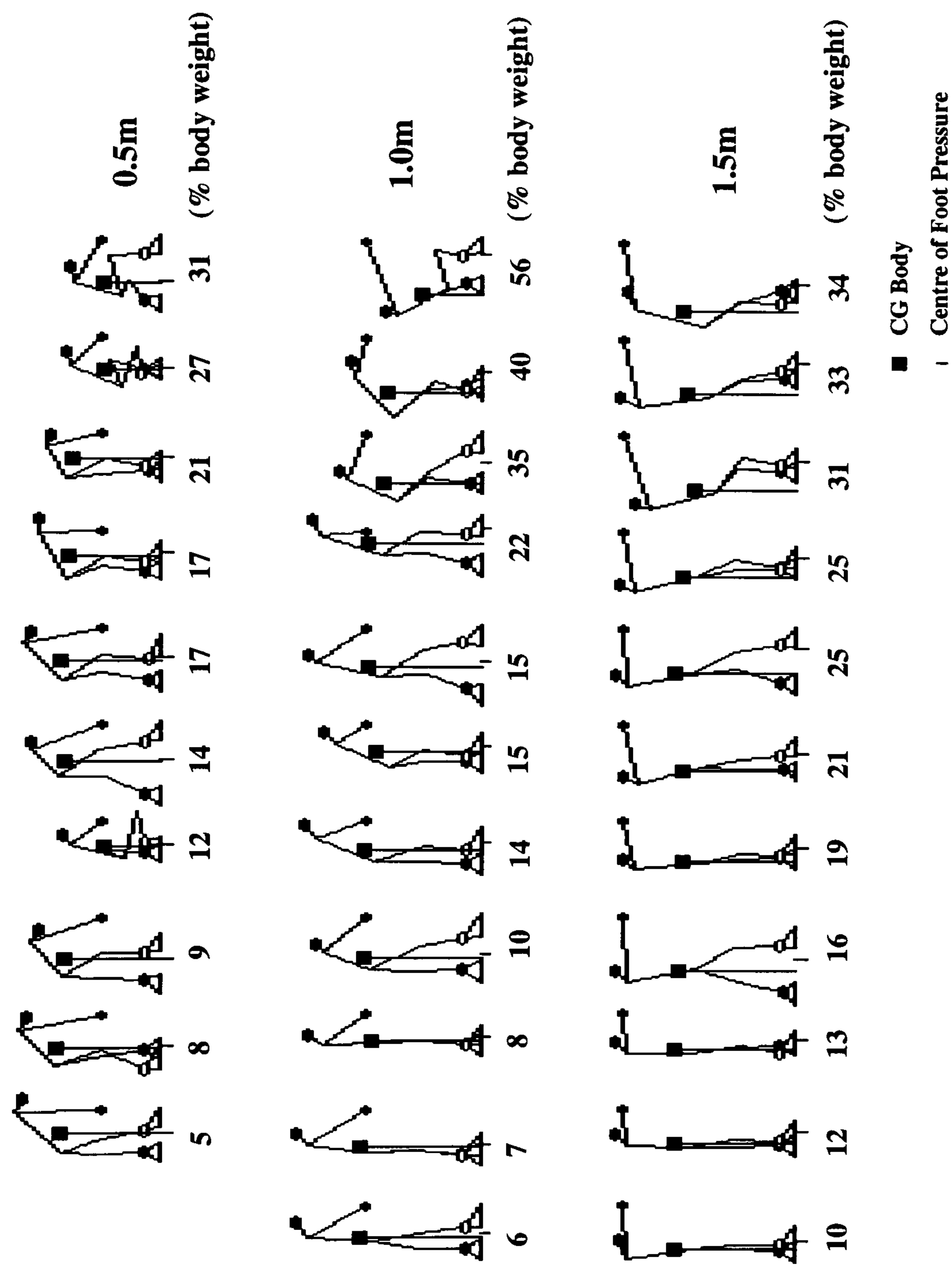


Figure 3.11: Postures of all subjects exerting down, backward and to the left at the three handle heights, showing forces achieved as a percentage body weight.

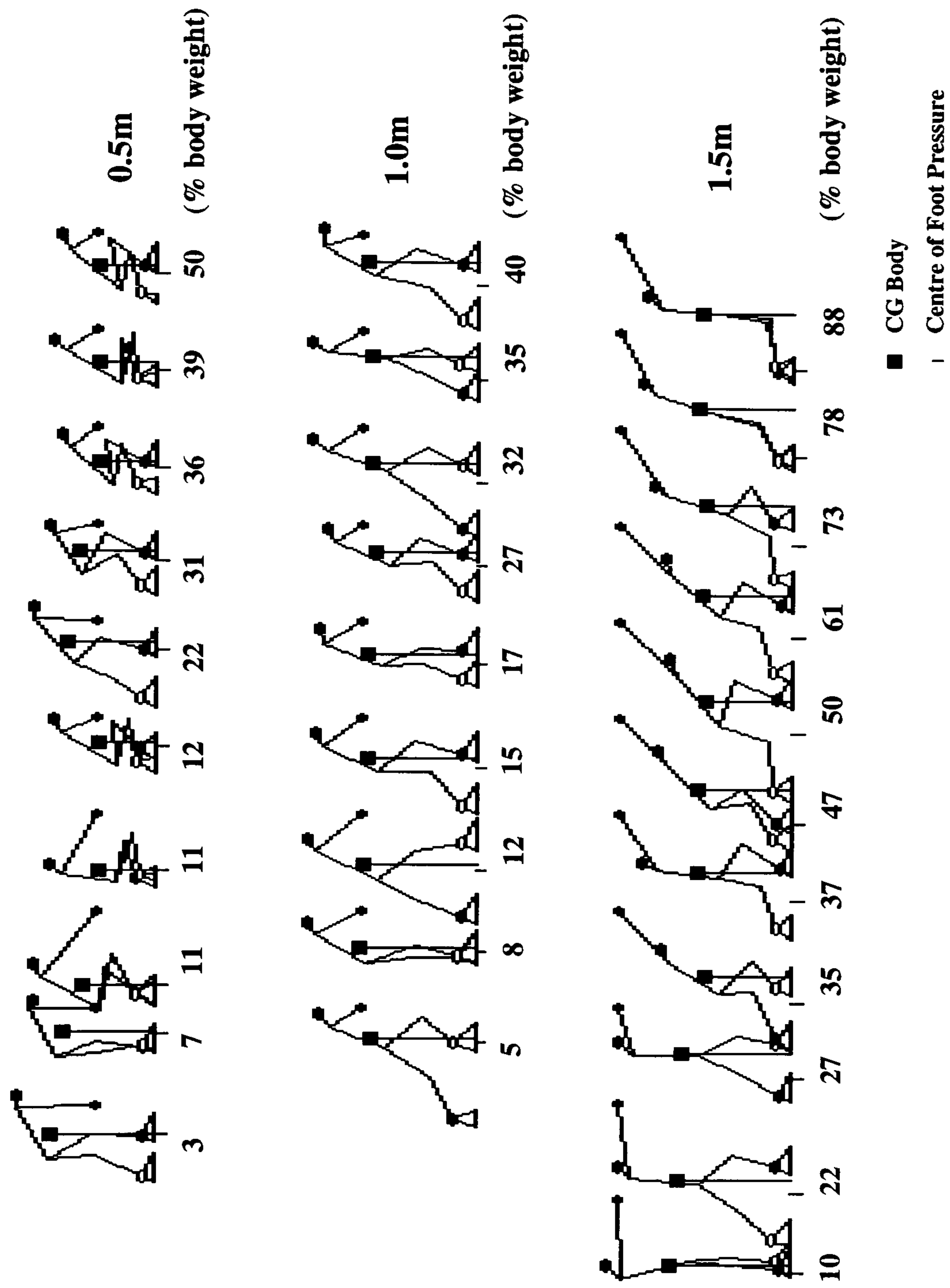


Figure 3.12: Postures of all subjects exerting up and forward at the three handle heights, showing forces achieved as a percentage body weight.

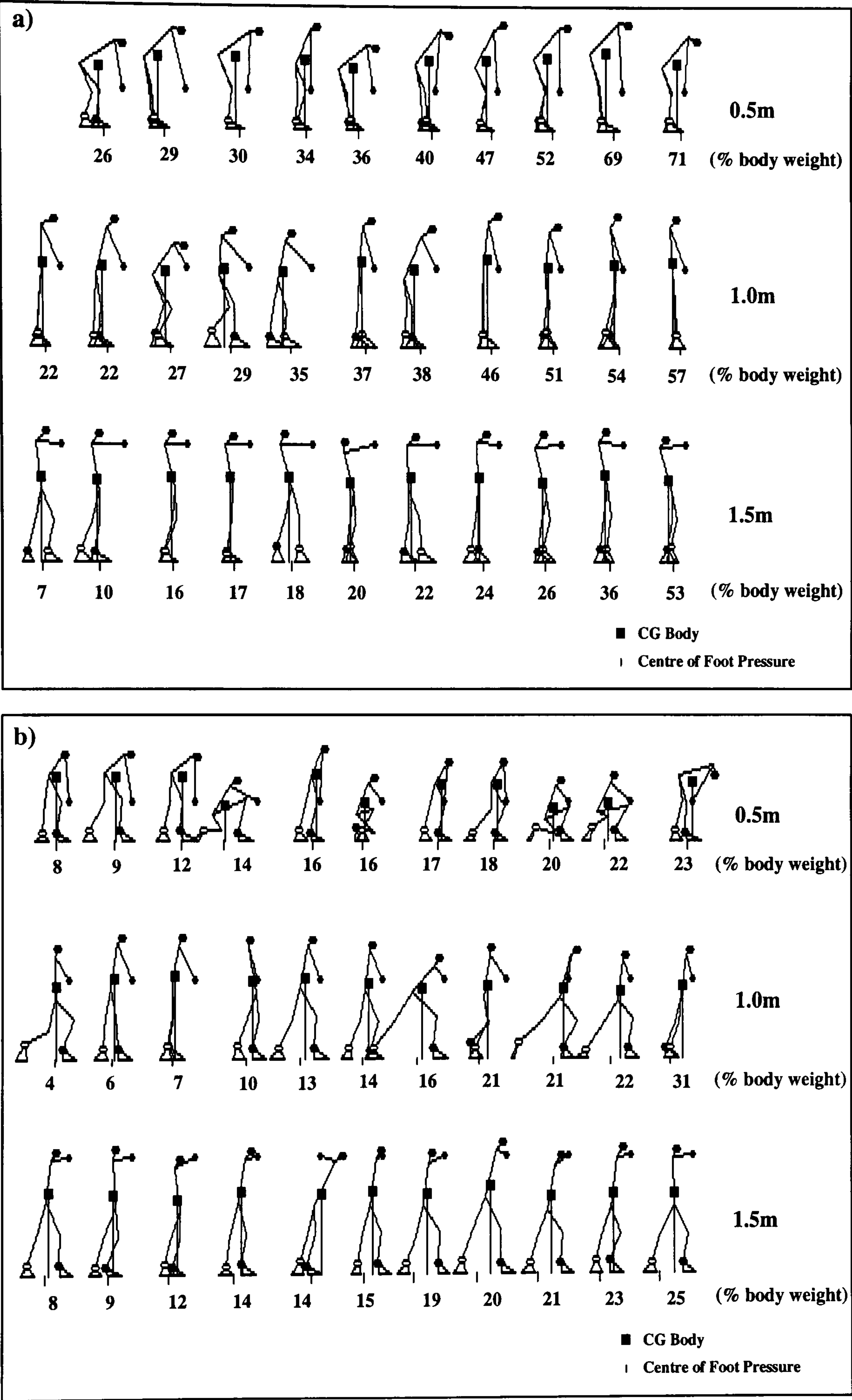


Figure 3.13: Postures of subjects exerting a) vertically upward, and
b) upward, forward and to the left.

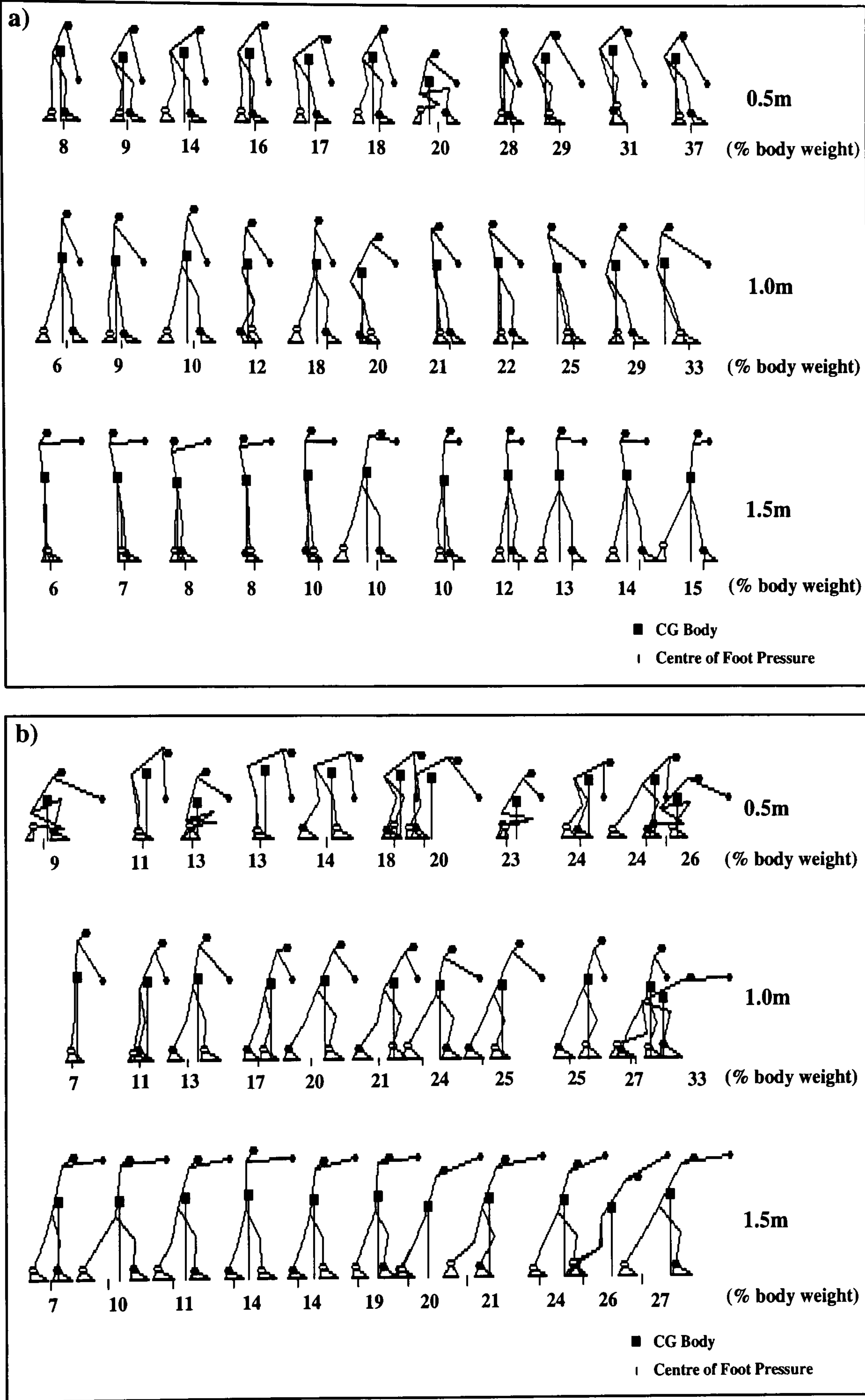


Figure 3.14: Postures of subjects exerting a) upward, backward and to the right, and b) horizontally forward.

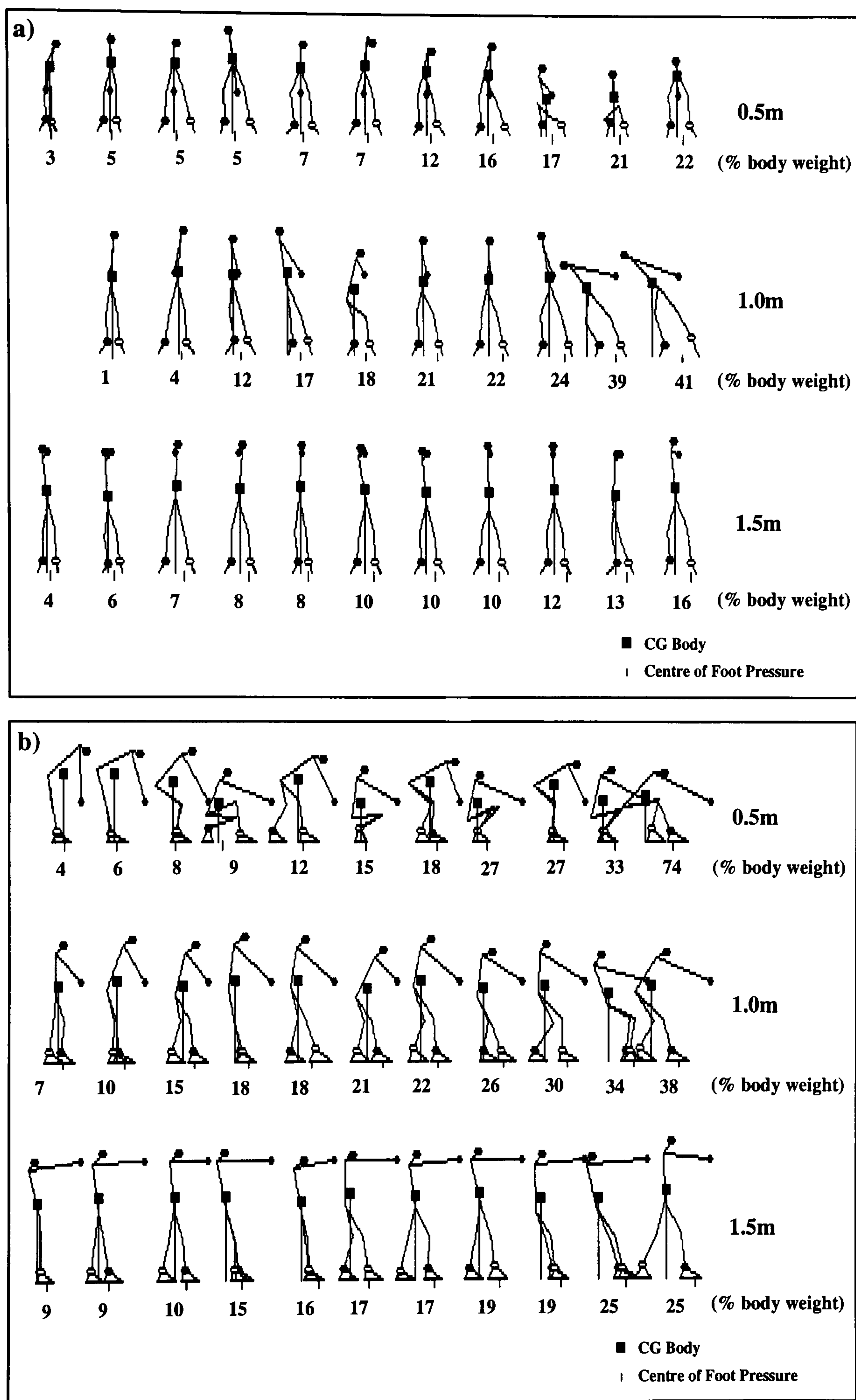


Figure 3.15: Postures of subjects exerting a) horizontally to the left, and b) horizontally backward.

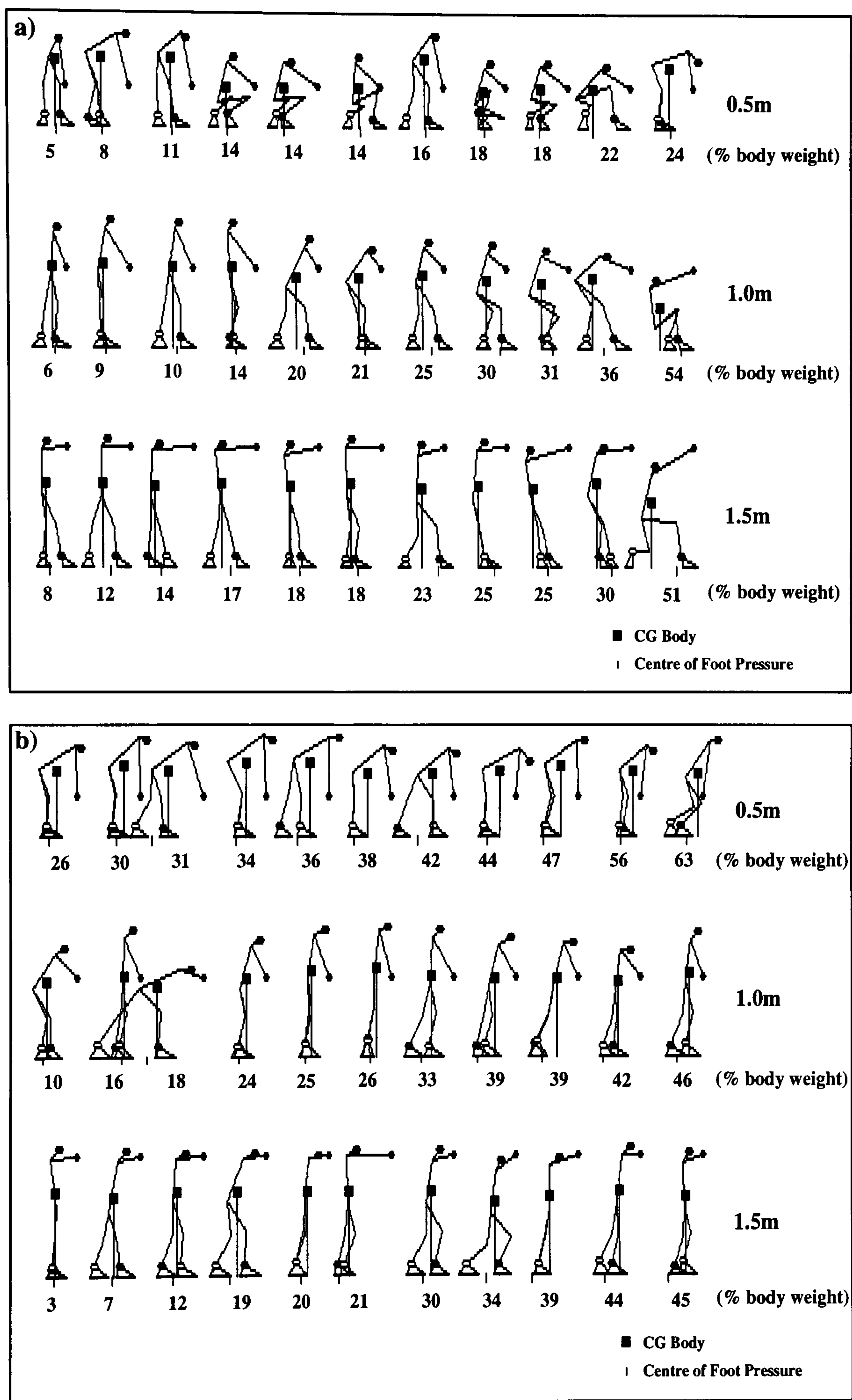


Figure 3.16: Postures of subjects exerting a) down, backward and to the right, and b) vertically downward.

directions are given in Appendix A.

3.3.2 *Live and dead components.*

Figure 3.17 shows the mean percentage of *live* force component for all directions at the three handle heights. Exertions with a large horizontal component were composed mostly of *dead* force, while exertions with an upward or downward component consisted of a greater proportion of *live* force. The first group of conditions rely more on the deployment of body weight, whereas it is difficult to benefit from body weight deployment when lifting vertically upward, for example. The patterns for heights of 1.0m and 1.5m are similar, while that for 0.5m differs slightly, in that there is a greater percentage of *live* force in directions with both a horizontal and backward component and a lesser percentage *live* force with an upward and backward component. This is not surprising, because at the higher handle heights, the subject can pull horizontally on the handle by leaning backwards, whereas with the handle at 0.5m, it is too close to the ground for this to be accomplished easily. On the other hand, it is easier for the subject to pull upwards on the handle when it is close to the ground.

It is surprising that pulling backward and down at handle heights 1.0m and 1.5m consists largely of *live* force. It would be expected that at these heights, the subjects would have been able to deploy their body weight to greater effect.

However, the percentage of *live* force shown is the mean of all eleven subjects.

Figure 3.18 shows the posture of the subjects with the weakest and strongest

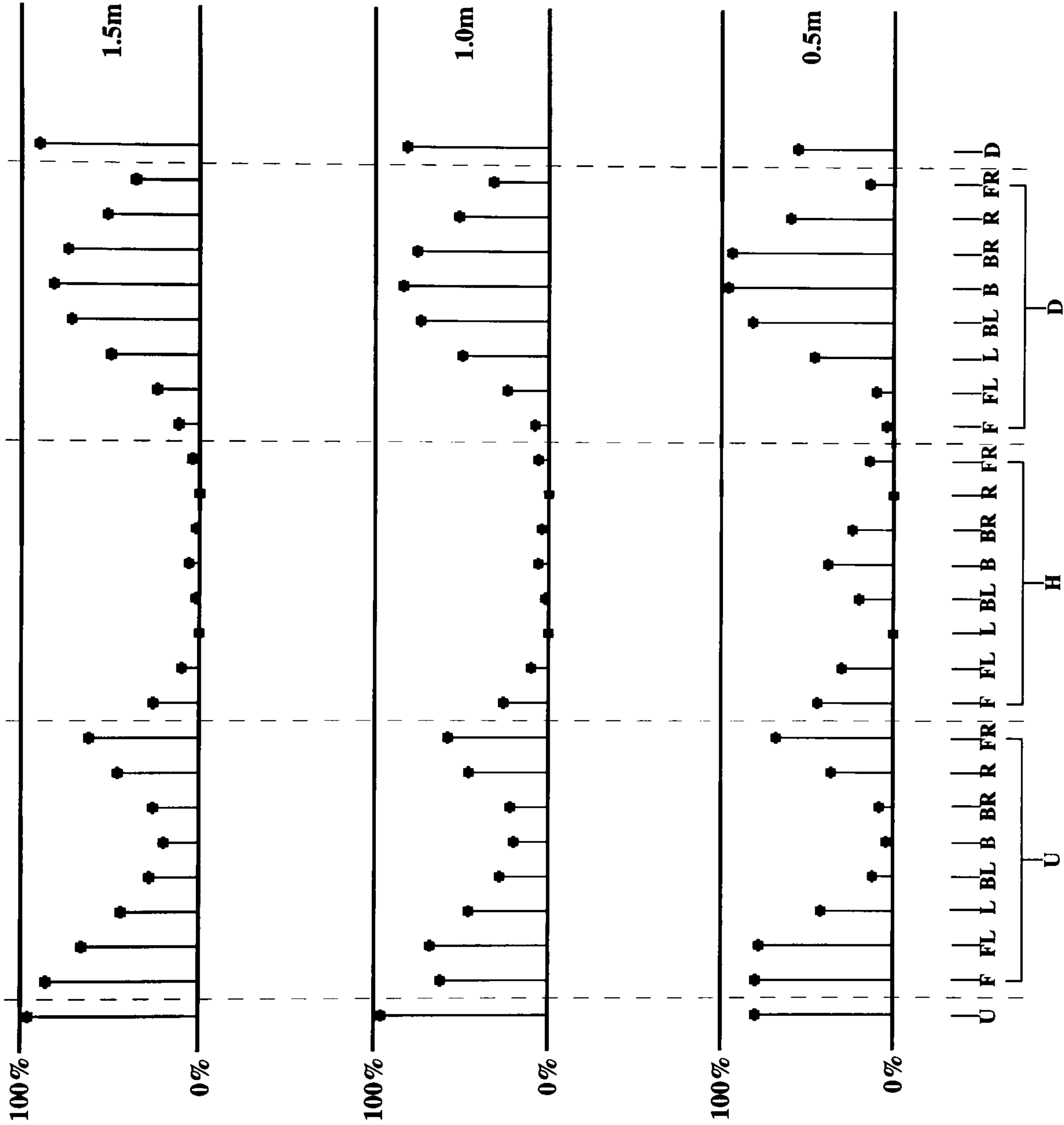
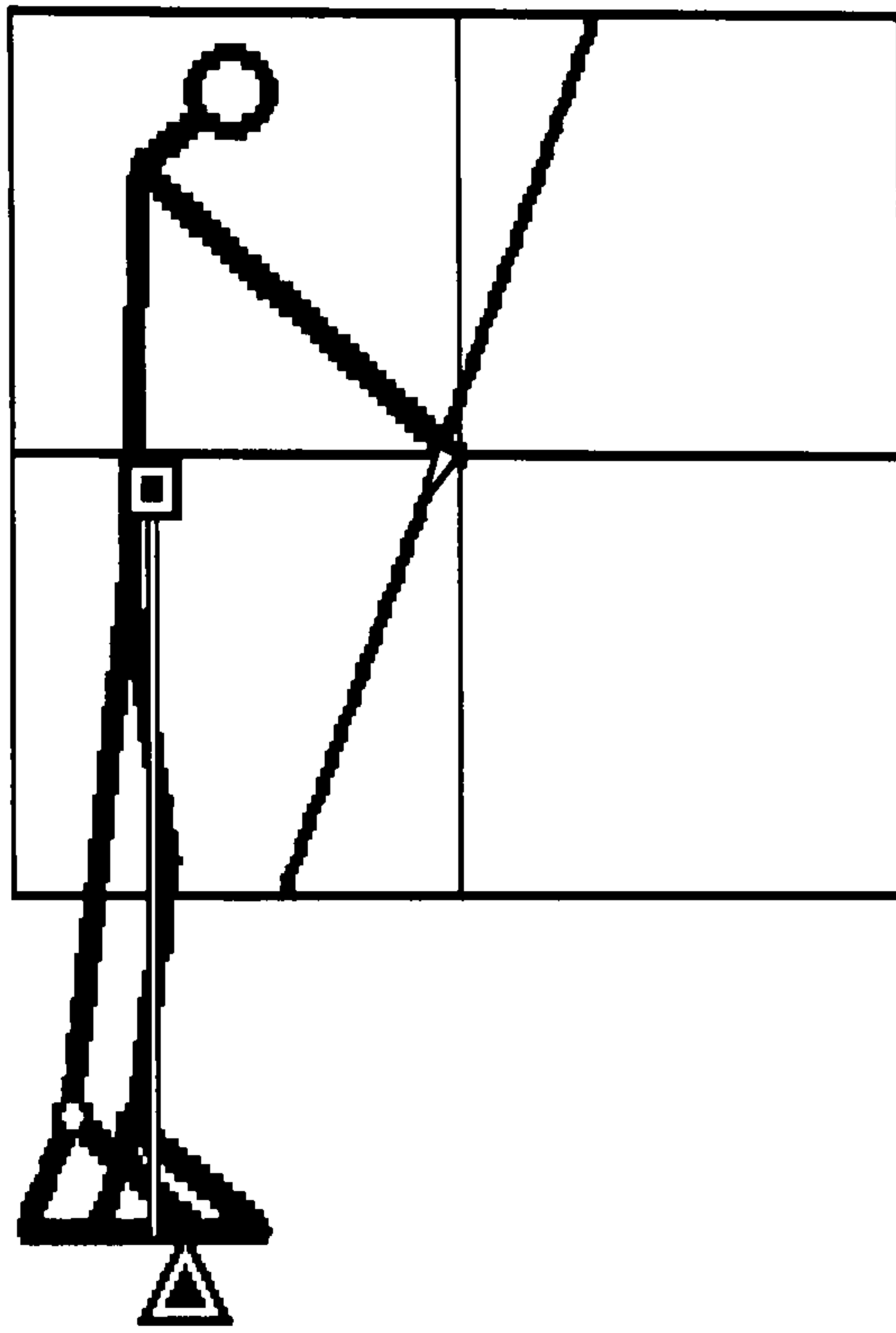
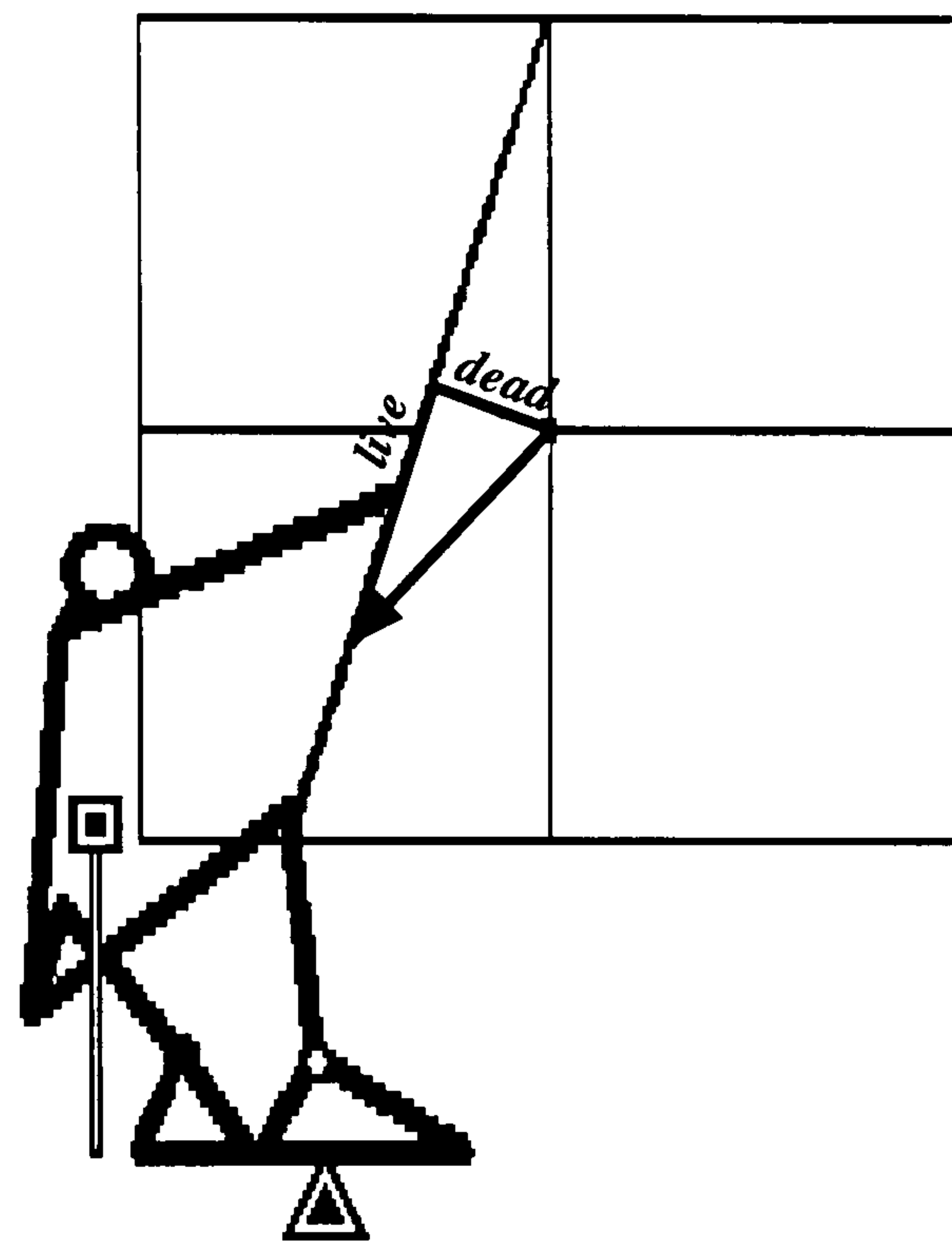


Figure 3.17: The mean percentage of *live* component of force in all twenty-six directions at the three handle heights. U - up, D - down, L - left, R - right, F - forward, B - backward.



a) Weakest exertion



b) Strongest exertion

Figure 3.18: The subjects with a) the weakest, and b) the strongest exertion pulling backward and down at handle height 1m. The Postural Stability Diagram is superimposed and shows the equation of static exertion with the *live* and *dead* components of the force vector.

exertions at handle height 1.0m superimposed on a Postural Stability Diagram. The *live* (along the equation of static exertion) and *dead* (at right-angles to the equation of static exertion) components of force are also shown. It is obvious that the subject with the stronger exertion is deploying his body weight to the full, whereas the weaker subject is standing very upright. The fact that many of the subjects chose not to employ body weight is probably the reason why the mean *dead* component is so small in a force direction where it would be expected to be high. Pushing forward and down at the same heights has a much larger proportion of *dead* force. In these cases, the subjects did not need to change their posture so dramatically and therefore a greater proportion of the force is *dead* component. At 0.5m, the force is almost entirely composed of *dead* force, as the subjects can lean forward and down on to the handle.

3.3.3 Moments about the feet.

Figure 3.19 shows posterior and superior views of the subject who exerted the greatest force horizontally and to the right at handle height 1.5m. Because the exertion was static, the force direction included a left or right component, and the hand was located anterior to the centre of foot pressure (Figure 3.19b), a turning moment at the feet in the horizontal plane was required to counterbalance the force at the hands.

Moments about the centre of foot pressure in the horizontal plane were calculated for each height, position and force direction. Figure 3.20 shows the mean moment

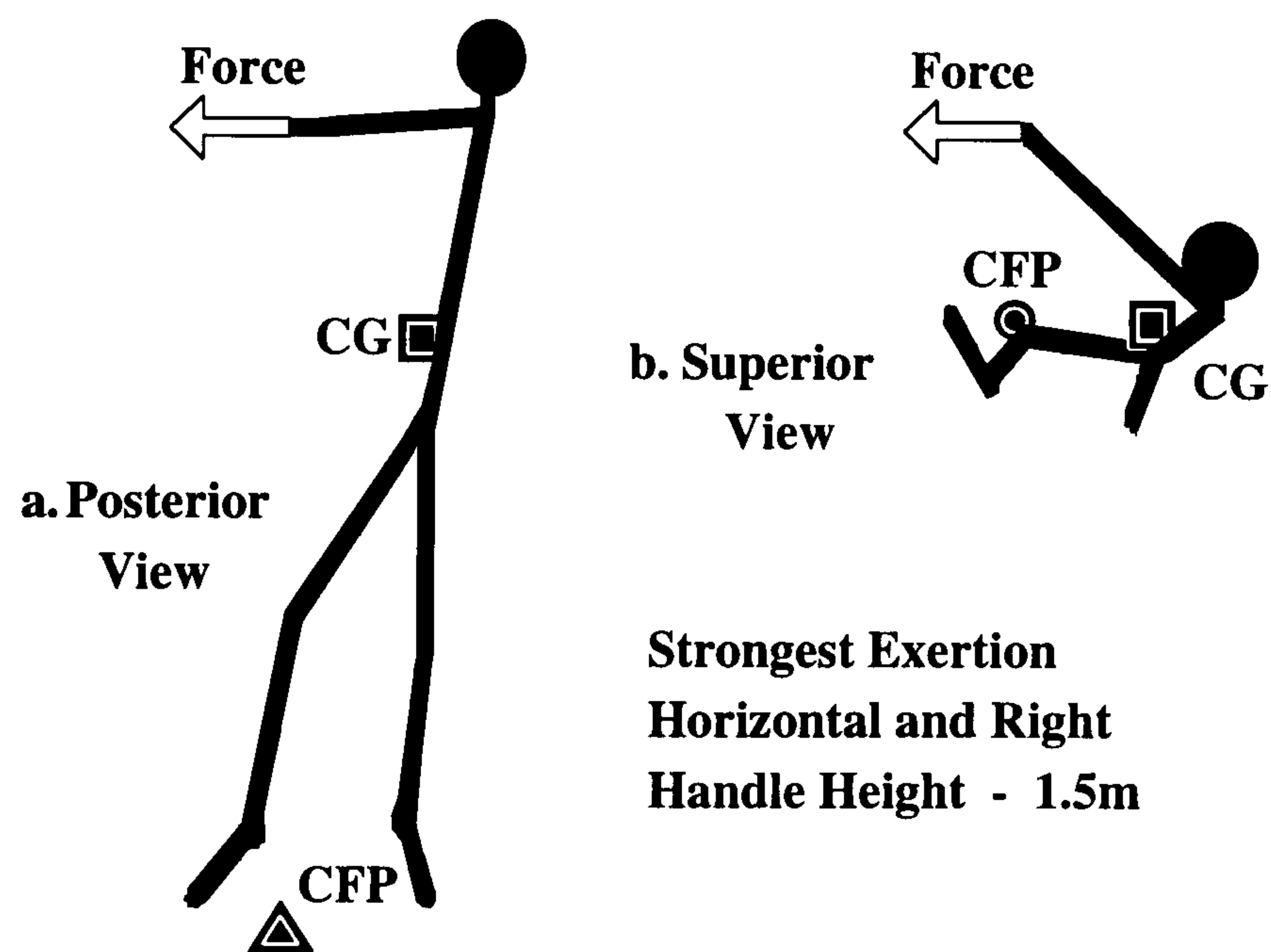


Figure 3.19: a) Posterior and b) superior views of the subject exerting with greatest force horizontally and to the right at handle height 1.5m.

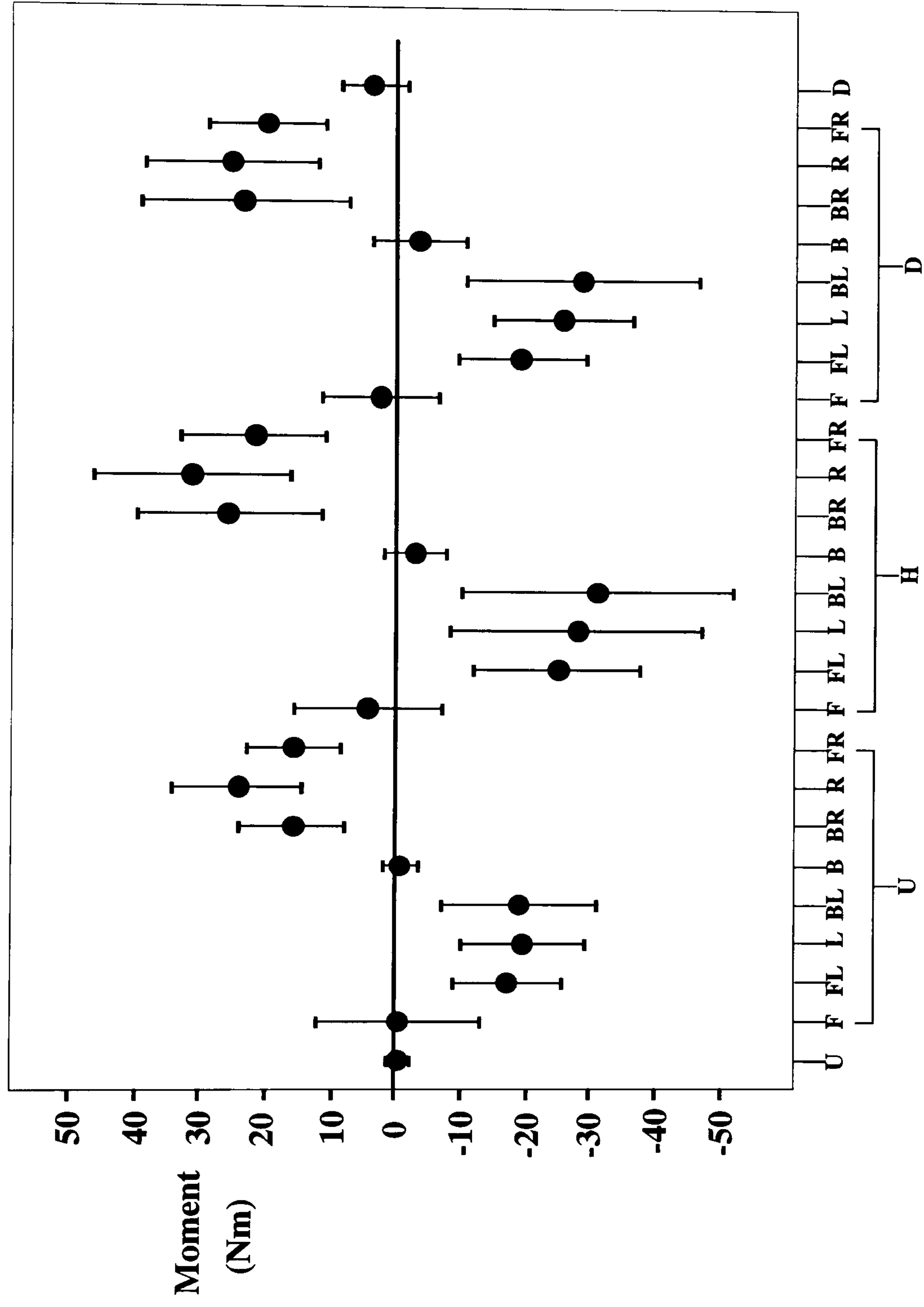


Figure 3.20: Moment (mean and standard deviation) upon the feet in the horizontal plane (positive clockwise when viewed from above). U - up, D - down, L - left, R - right, F - forward, B - backward.

upon the feet, with standard deviation, for all three heights and twenty-six force directions. The data for the three heights were combined, because there was no significant influence of height on the magnitudes of the calculated moments at each height (ANOVA: F-ratio = 0.01, $p = 0.988$).

The difference between the magnitude during force exertion with a left or right component and that during exertion in the vertical fore-aft plane is clearly illustrated. In the latter case, whether a value was clockwise or anti-clockwise depended upon the exact direction of the exerted force, which was allowed to deviate up to 5° from the target direction. Forces with a left or right directional component were always accompanied by mean horizontal moments at the feet greater than those of forces in the fore-aft plane. The largest moments (mean plus one standard deviation) reached magnitudes of 50Nm.

3.3.4 Postural Stability Diagram.

If the co-ordinates of the landmarks, the CG and the centre of foot pressure are projected into the vertical plane containing the force vector, which may or may not be in the fore-aft plane, a Postural Stability Diagram can be drawn and the equations of static exertion implemented. Figure 3.21 shows an example of the Diagram applied to the subject with the greatest force when exerting up and to the left at handle height 0.5m. Overlaid on the stick figure is the outline of the Postural Stability Diagram. As before, the centre of the square represents zero forces at the hands, and force vectors may be drawn from this origin showing the

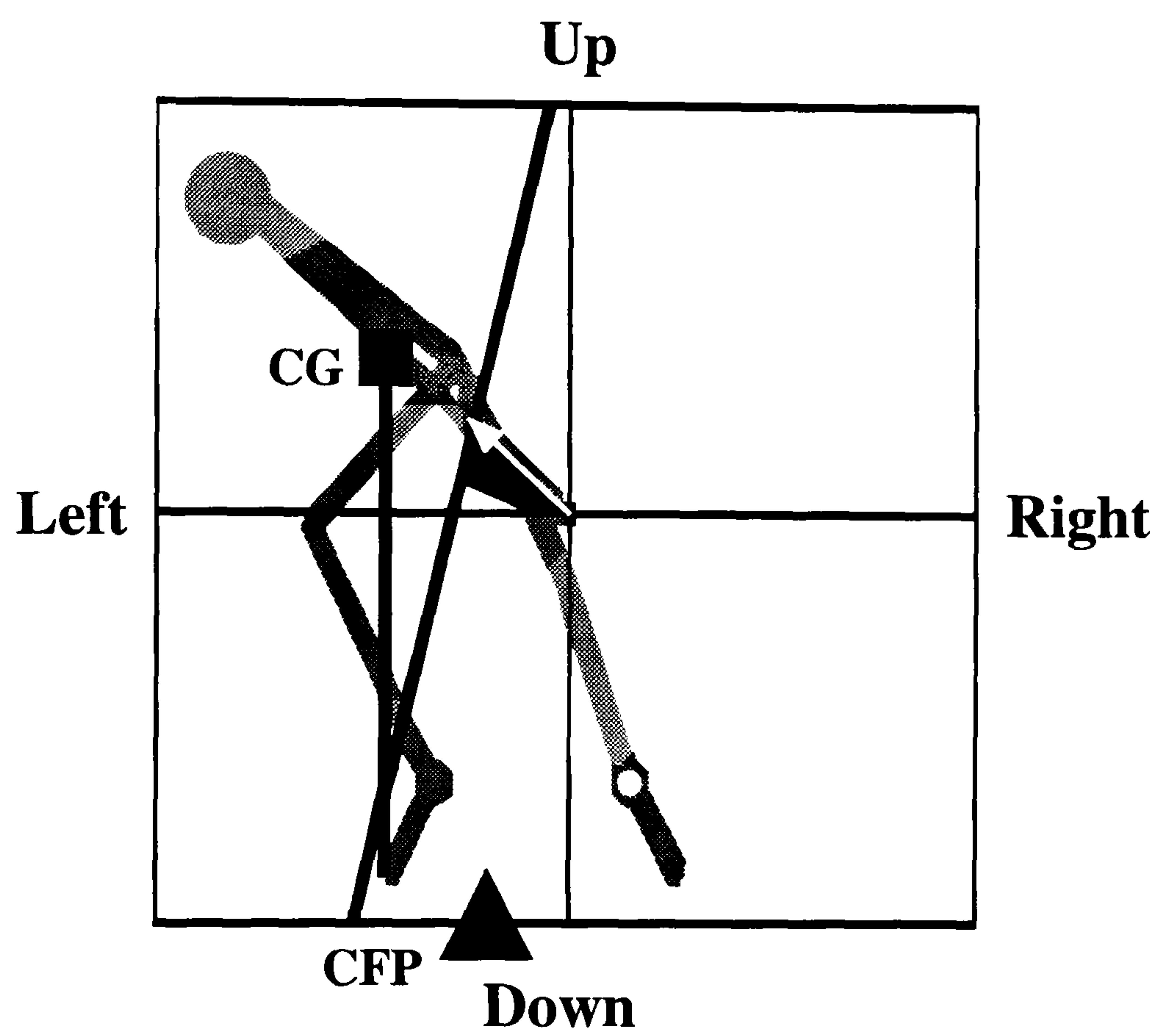


Figure 3.21: Postural Stability Diagram (subject exerting up and to the left at handle height 0.5m).

force exerted by the subject as a percentage of body weight. The line of the equation of static exertion is also shown. Making a triangle with the force vector, the two sides along the line of the equation of static exertion and at right-angles to it, represent the *live* and *dead* force respectively.

3.4 DISCUSSION.

Because the subject's posture was recorded as soon as the desired direction had been achieved, the force exerted at the moment the photograph was taken was a variable fraction of the strength which had been determined for that direction during the previous four days. It was therefore possible to compare and contrast postures associated with weaker and stronger exertions.

The subjects who exhibited greater postural mobility succeeded in exerting greater forces than those who changed their posture relatively little. Because the *dead* force component is due to the deployment of body weight, this mobility was more important in force directions where the *dead* component made up a large percentage of the exerted force (e.g., horizontally to the right or left) and less so where the *live* component was the greater (e.g., vertically upward). However, the greater forces attained by some subjects in directions where the *dead* component was substantial, were achieved by an increase in both *dead* and *live* component. This suggests that subjects who achieved greater forces did so by both increased effort and better deployment of body weight.

It is recognised that the positions of the CG and centre of foot pressure were calculated rather than measured and that the CG of the upper limbs was based on an assumed position. However, these assumptions were considered to be reasonable because in the context of the whole body, the upper limbs make up only 10% of the body weight (Dempster, 1955), whereas the CG's of the large body parts such as the torso and thighs were known more accurately. The calculated errors in the position of the CG and centre of foot pressure of the body introduced by assuming exaggerated postures of the upper limbs were only 39mm and 38mm respectively. This indicates that errors associated with the actual assumptions made would be acceptable.

In this study, the use of the Postural Stability Diagram has been extended into any vertical plane which includes the manual force vector. This planar analysis simplifies the consideration of the three-dimensional reality. The Postural Stability Diagram now allows the prediction of possible combinations of vertical and horizontal forces in any situation of static exertion where the centre of foot pressure, CG and point of application of force are known. Projecting these points into the plane containing the force vector allows the equation:

$$\frac{UP}{W} = \frac{HORIZ}{W} \cdot \frac{h}{b'} - \frac{a'}{b'} \quad (v)$$

to be used in each plane (where a' and b' are the horizontal distances between the centre of foot pressure and the CG of the body and between the centre of foot pressure and the centroid of force at the hands respectively, in the plane being considered). Again, drawing the line of the equation of static exertion at its

extremes on the Diagram will indicate the limits of any force that can be exerted with a particular foot and hand configuration.

In an industrial situation, knowing the limits of the exerable force due to deployment of body weight may be of use in training so that workers may be made aware of how a change in posture may improve their ability to carry out a task. Many of the subjects in these experiments chose not to employ body weight while exerting forces. These were almost invariably the subjects who exerted the least forces, particularly in directions where the *dead* component of force was an important factor. This demonstrates how training could be important.

When left or right components exist in manual exertion, the horizontal moment at the feet is apparently an important factor in the demands made on the body. The largest moments (mean plus one standard deviation = 50Nm) were found with a horizontal exertion to the left at handle height 1.0m. These must be transmitted by the structures of the lower limb. There are various ways in which this might be achieved. Medial and lateral rotators could generate a turning moment in the plane of the floor. Simultaneous employment of the hip extensors and knee flexors of one lower limb and the hip flexors and knee extensors of the other would also achieve a turning moment. The strategies adopted for simultaneous exertion of turning moment and force are not understood and will be investigated in the next chapter.

3.5 CONCLUSIONS.

1. The results suggest that more effective use of body weight as well as increased effort was employed by subjects who exerted greater forces.
2. The Postural Stability Diagram, together with the equation of static exertion, may be used in the consideration of forces exerted in directions other than those in the fore-aft plane.
3. Exerting in directions with a left or right component introduces a new finding of a horizontal moment at the feet, which is independent of handle height and may reach magnitudes as great as 50Nm. This moment is an important factor in the demands made upon the body during asymmetrical exertion.

CHAPTER 4.

**USE OF SURFACE ELECTROMYOGRAPHY AND A FORCE
PLATE TO INVESTIGATE THE GENERATION OF BOTH
HORIZONTAL TORQUE AND THRUST AT THE FOOT-BASE.**

4.1 INTRODUCTION.

In the previous chapter, the existence of a large horizontal moment at the feet during exertions to the left or right was deduced. Mechanisms by which the turning moment is generated may be suggested:

Mechanism A: one foot may exert a forward (or left) directed force on the floor at the same time that the other exerts a backward (or right) directed force, causing a turning moment;

Mechanism B: each foot may exert an individual torque upon the floor.

With Mechanism A, the lateral forces seem less likely for two reasons. Firstly, one foot must be in front of the other for a moment to be produced, whereas the feet of subjects in our previous experiment were frequently aligned in the coronal plane. Secondly, the laterally directed force at the hands must be counteracted by one at the feet of equal magnitude, but opposite in direction. Since the hand forces were often large, the sum of laterally directed force vectors at the foot-base must also have been large, which would be difficult to achieve if the feet were applying forces to the floor in opposite directions.

The demands on the muscles of the lower limb during exertions with a lateral directional component are therefore not only to maintain the posture and support the body weight, but also to counteract the lateral force at the hands by applying a laterally directed thrust on the floor and generating a horizontal moment.

Anatomically, the muscles of the lower limb must interact in a complex manner for the resulting forces and moments to be applied simultaneously during lateral exertions.

4.1.1 Exertion to the left.

The actions of the muscles of the right lower limb (and therefore the left lower limb when exerting in the opposite direction) may be considered during lateral force production at the right hand, beginning with exertion to the left.

When a left-directed force is exerted on a handle, the counteracting moment exerted by the feet upon the floor is clockwise when viewed from above. The counteracting force upon the floor is to the right, which suggests hip abductor activity in the case of the right lower limb. Presuming the existence of mechanism A, which in this case would be achieved by the left foot pushing forward and the right foot pushing backward on the floor, the expected muscle activity in the right lower limb would be that of the hip extensors and knee flexors. Mechanism B (each foot exerting an individual torque on the floor) would be achieved with activity of the lateral rotators.

4.1.1.1 *The foot and ankle.*

It is likely that during lateral exertions, the structures of the foot (bones, ligaments, extrinsic and intrinsic muscles) serve to transmit forces and torques that have been

generated by stronger proximal muscles.

4.1.1.2 *The knee.*

Mechanism B requires lateral rotator muscle activity. The range of rotation at the knee can be large, depending on the posture of both the knee (flexed) and the hip (Shoemaker and Markolf, 1982). Since most of the subjects in the previous experiment exerted with a flexed knee, biceps femoris could have provided lateral rotator activity. The lateral rotatory torque at the knee is usually weaker than medial torque and has been measured to be 20.0Nm (s.d. 6.7) with 10° of hip flexion and 20° of knee flexion (Shoemaker and Markolf, 1982). This value increases significantly (to 25.4Nm) if the knee is flexed to 90°.

As flexion of the knee is increased, the lateral rotators of the hip become less able to transmit the same moment to the leg. If the knee is flexed to an angle of 90°, lateral rotation (and, with the foot fixed on the ground, abduction) of the thigh would rather cause lateral angulation of the leg on the foot. If the knee is fully extended or close to fully extended, it would be possible for passive mechanisms simply to prevent rotation at the knee, so that the torque is transmitted to the tibia. Full extension of the knee is the close-packed position, where parts of both cruciate ligaments, the lateral and medial collateral ligaments, the posterior capsular region, oblique posterior ligament, skin and fasciae are all taut (Gray, 1989), and the tibial intercondylar eminence and femoral intercondylar notch are locked together. All these structures would resist the applied torque, because the only possible rotation

of the tibia in this position under normal circumstances is medial rotation as the joint "unlocks" at the beginning of knee flexion.

Mechanism A would require flexor activity at the knee during exertion to the left, which could be provided by the hamstrings and gastrocnemius. With the knee in full extension, the backward force at the foot-base could be achieved by extensor activity at the hip, unless active mechanisms at the knee are required to resist hyperextension. However, the knees of the subjects in the previous experiment were more commonly flexed during this exertion. To maintain this position of slight flexion, quadriceps femoris would be required.

4.1.1.3 *The hip.*

At the hip joint, activity would be expected in the extensors (Mechanism A), lateral rotators (Mechanism B) and abductors. Muscles normally responsible for extension are gluteus maximus, the hamstrings and the posterior part of adductor magnus; for abduction are gluteus minimus and medius, the upper fibres of gluteus maximus and tensor fasciae latae; and for lateral rotation are piriformis, gemellus superior and inferior, obturator internus and externus and quadratus femoris.

Rotation of the femur is greater in range with a flexed hip, and occurs around the mechanical axis (connecting the hip and knee joint centres) rather than the long axis of the bone (Palastanga *et al.* 1989), so that the greater trochanter and upper part of the femur move backwards as well as rotating. In postures other than the

anatomical position, the function of the muscles may change. Dostal *et al.* (1986), with straight line representations of lines of muscle action and a dry bone specimen, used the concept of a moment arm vector to describe the actions of muscles crossing the hip joint. The moment arm vector of a muscle is the product of the moment arm and a unit vector indicating the direction or rotational sense of the moment using the right-hand rule. It therefore has the same magnitude as, but a different direction to, the actual moment arm. Moment arm vectors were measured at 0°, 40° and 90° of hip flexion. Some actions of muscles decreased or increased depending on the position of the hip, and even the direction of some secondary muscle actions was reversed. For example, piriformis was always an abductor, but a lateral rotator at 0° and a medial rotator at 90° of hip flexion. Adductor longus was always an adductor, but a flexor from 0° to approximately 50° of hip flexion and an extensor from 50° to 90°. None of the muscles' major actions changed direction. It must be noted that the moment arm vector is based on geometric concepts rather than observed muscle activity.

Since the posture of subjects in Chapter 3 was generally one of slight hip flexion, the hamstrings would probably have been required for postural maintenance.

4.1.2 Exertion to the right.

When a force to the right is exerted on a handle, the counteracting moment exerted by the feet upon the floor is anti-clockwise when viewed from above, and the laterally directed force at the hands implies a counteracting force upon the floor to

the left, which suggests hip adductor activity in the right lower limb. Mechanism A suggests a forward directed force on the floor, which could be achieved by flexor activity at the hip and extensor activity at the knee, while Mechanism B would require activity of the medial rotators.

4.1.2.1 *The foot and ankle.*

Again, the foot probably serves to transmit forces and torques from proximal structures to the floor.

4.1.2.2 *The knee.*

Mechanism A would require extensor (quadriceps femoris) activity at the knee. Mechanism B could be achieved by the activity of the medial rotators (semimembranosus and semitendinosus) with the knee flexed. A medial rotator torque of 30.9Nm (s.d. 9.6) has been measured with the hip flexed to 10° and the knee flexed to 20° (Shoemaker and Markolf, 1982). With the knee extended, medial rotation of the thigh would be transmitted to the leg.

4.1.2.3 *The hip.*

The medial rotators of the hip are the anterior fibres of the gluteus medius and minimus and the tensor fasciae latae. Other muscles involved in medial rotation are controversial. Dostal *et al.* (1986) and Palastanga *et al.* (1989) list iliacus and

psoas as medial rotators of the hip, although the latter admit that controversy still exists, whereas Basmajian (1974) decides in favour of both iliacus and psoas as weak lateral rotators after experiments using long needle and fine wire electrodes in electromyographic studies. Gray's Anatomy (1989) agrees that psoas is probably a lateral rotator. Muscles with lines of action passing in front of the mechanical axis will cause medial rotation, while those passing behind will induce lateral rotation (Palastanga *et al.* 1989), but the relative positions of the pelvis and femur may well change the action of the muscle. In exertion to the right, if Mechanism A is valid, iliopsoas would in any case be providing flexor activity at the hip, together with pectineus, sartorius and rectus femoris, the last also assisting in extension of the knee. Adductor longus, brevis and magnus would also be expected to be active, together with gracilis and pectineus.

As the posture of subjects was generally that of flexion of the hip, the hamstrings may help support the weight of the upper body.

4.1.3 Experimental design.

Although Mechanisms A and B may be separated in terms of the actions of each lower limb upon the floor (a torque exerted by one limb on the floor is measurably different to a forward or backward force, and it is obvious that Mechanism A cannot exist if only one foot is in contact with the ground), anatomically they are not easily distinguished. For example, the deep lateral rotators of the hip may have an additional extensor role, depending on the position of the hip joint. According

to Dostal *et al.* (1986), the quadratus femoris is a lateral rotator (Mechanism B) at all positions of the hip from 0° to 90° of flexion, but is also an extensor (Mechanism A) at 40° and 90° and adducts the thigh at 0° of flexion. Biceps femoris is an extensor of the hip and flexor of the knee (Mechanism A), but it may also be a lateral rotator of the leg (Mechanism B) and help support the weight of the upper body when the torso is flexed on the thigh. It may therefore be difficult to state that the activity of any one muscle must be implicated in only one role during lateral exertions.

In an attempt to clarify some of the questions that have been raised by the above discussion, two experiments using the available facilities were designed. The first used surface electromyography to investigate the major extensors and flexors of the hip, knee and ankle during exertion of forces in four different directions in the horizontal plane. These muscles were chosen because of their likely involvement in Mechanism A and their easy accessibility. Investigation of another group of muscles of particular interest, the deep lateral rotators of the hip, would require the invasive procedure of fine-wire electromyography, which, although an attractive method of investigation, was not available. It would in any case have been unsafe to use on these muscles, because of the proximity of other structures. The second experiment used a force plate to measure the forces and moments at the foot-base by each foot individually and by both together during similar exertions. In the series of experiments in the previous chapter, the horizontal moment was calculated rather than measured. A force plate could provide real measurements of this

moment and be used to investigate the individual contributions of each lower limb to the moments and translational forces generated at the foot-base simultaneously.

In both experiments, measurements were taken while subjects exerted forward, backward, to the left and to the right on a handle placed at 1m above the ground. Other heights were not considered necessary, because in the previous set of experiments, no significant differences were found between horizontal moments calculated at the three different heights.

4.1.4 Aims.

The aims of the experiment were therefore:

1. to investigate, using surface electromyography, the role of the quadriceps femoris, hamstrings, tibialis anterior and gastrocnemius in the generation of both thrust and horizontal torque at the foot-base during laterally directed exertions at the hands;
2. to investigate the mechanisms by which the lower limb generates a horizontal moment during laterally directed force exertion at the hands;
3. to attempt to quantify the contribution of these mechanisms to the generation of horizontal torque;
4. to measure the magnitude of the horizontal torque.

4.2 SURFACE ELECTROMYOGRAPHY - MATERIALS AND METHODS.

4.2.1 Experimental equipment.

Electromyographic (emg) readings from four muscle groups of the right lower limb were taken while subjects stood on a platform and exerted forces on the handle of the Tri-Axial Force Measurement System (TAFS) (Figure 4.1). The platform, which was divided into right and left halves by a central, longitudinal line, measured 1.6 x 0.95m and was covered with emery paper. The TAFS handle was mounted 1m above the central line at the front of the platform, and a computer monitor above and behind the handle provided feedback. A video camera on a tripod was positioned to the right of the subject at a distance of 4m, with its axis approximately horizontal and perpendicular to the fore-aft plane.

4.2.1.1 *Electromyography (emg).*

Surface electrodes were placed on the skin over four muscle groups (the quadriceps, hamstrings, tibialis anterior and gastrocnemius), using the positions given in Basmajian and Blumenstein (1980) (Figure 4.2). Electrode sites were shaved, scrubbed with an abrasive pad, rinsed, dried and swabbed with alcohol. Pairs of cupped, 10mm diameter silver-silver chloride surface electrodes were applied with double-sided tape 15 to 20mm apart. The electrodes were filled with

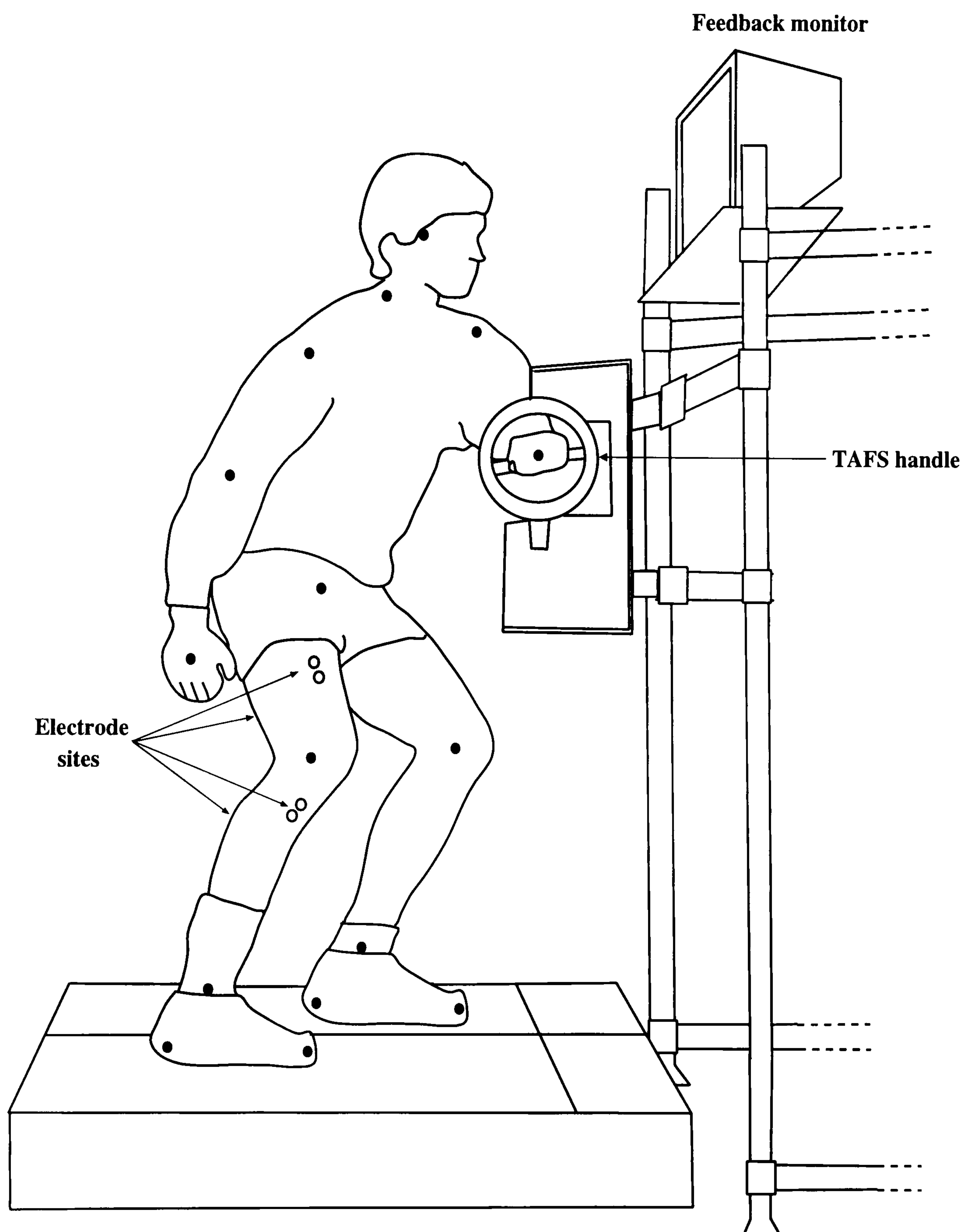


Figure 4.1: Experimental set-up, showing digitized points.

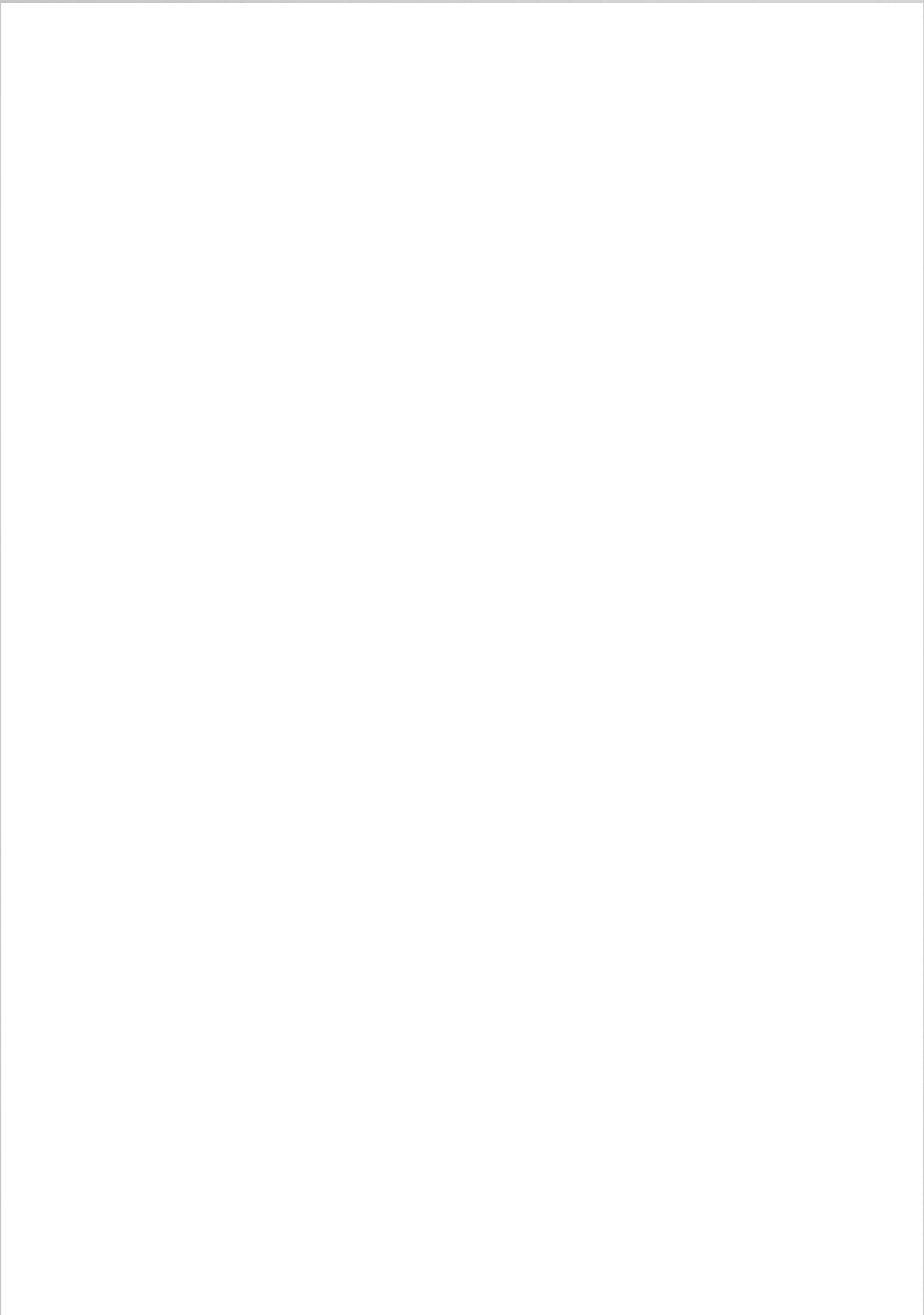


Figure 4.2: Placement of surface electrodes (from Basmajian and Blumenstein 1980).

jelly by means of a syringe and blunted needle, and their leads taped to the skin with Micropore tape to prevent electrode dislodgement. The ground electrode was placed on the right ankle. The emg signals passed through pre-amplifiers on the back of a waist-belt to four AA6 Mk II amplifiers in a Medelec MS6 Modular Electrophysiological System. The amplifier had a bandwidth (3dB points) of 32 to 800Hz with a gain setting of 50 μ V/division. The signals were then rectified and smoothed by a leaky integrator (time constant approximately 0.1s) and sampled at 500Hz by analogue-to-digital channels of a CED1401. The readings were transferred to the host computer (Archimedes A310) and the mean emg values over a four second period obtained (for each muscle group, exertion and subject).

4.2.2 Subjects.

Ten male subjects of mean age 28yrs (s.d. 4.7), height 1.77m (s.d. 0.66) and weight 75.9kg (s.d. 10.5) took part in the experiment.

4.2.3 Experimental procedure.

Each subject participated on two separate days. On both days, the subject exerted forces on the TAFS handle horizontally to the right, left, backward and forward with either the left or right hand in six different combinations, given in Table 4.1.

Table 4.1: The six combinations of direction of exertion and hand used.

DIRECTION	HAND
Forward	Right
Backward	Right
Left	Left
Right	Right
Right	Left
Left	Right

4.2.3.1 *Day 1*

The purpose of the first day was to familiarize the subject with the apparatus and procedure and to measure the maximum force for each of the six conditions. A consent form was signed by all subjects.

The subject stood on the platform with both feet facing forwards (Figure 4.1), parallel to and equidistant from the central line, but not necessarily opposite each other. A short practice session followed to enable the subject to become accustomed to exerting forces on the handle in different directions and to familiarize himself with the feedback displayed on the computer monitor. This consisted of a circle (representing the horizontal plane), with a sector marked to indicate the required horizontal direction (Figure 4.3). The radius of the circle was equivalent to 100% of the subject’s body weight. On the right of the screen was a

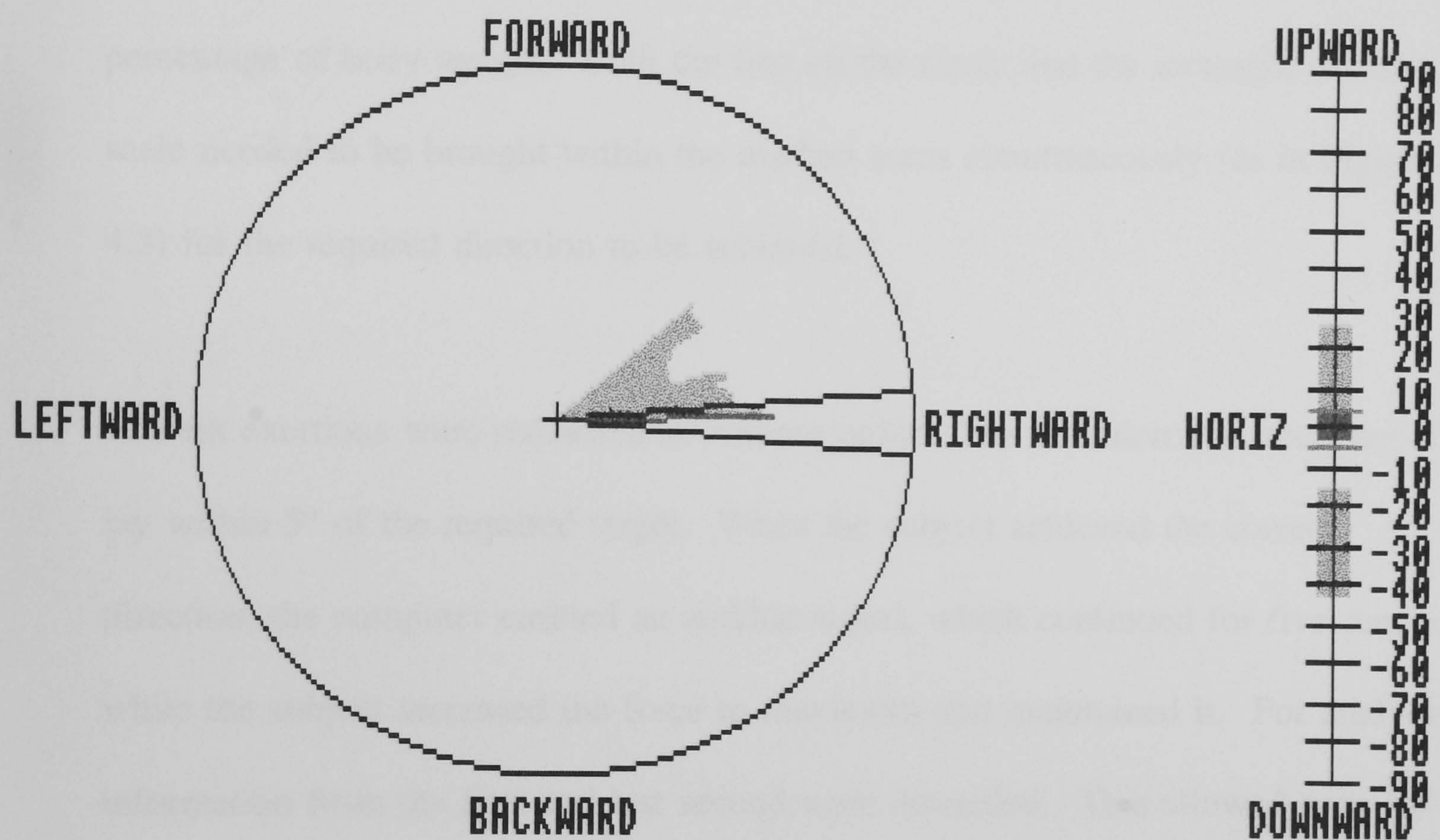


Figure 4.3: Feedback displayed on computer monitor. The subject has just achieved the required force direction of horizontal and to the right.

scale representing the vertical axis. This consisted of a vertical line, the middle of which represented horizontal, with a scale marking 5° and then every 10° above and below this. As the subject exerted in different directions, a red trace appeared instantly on both the circle and vertical scale, with a yellow shadow indicating the directions already attempted. On the circle, the trace appeared as a line radiating out from the centre, and on the vertical scale as a small solid rectangle. The length of the radiating line on the circle represented the force of the subject as a percentage of body weight. Both the line on the circle and the rectangle on the scale needed to be brought within the marked areas simultaneously (as in Figure 4.3) for the required direction to be achieved.

The six exertions were requested in random order. The direction was accepted if it lay within 5° of the required target. When the subject achieved the correct direction, the computer emitted an audible signal, which continued for five seconds while the subject increased the force to maximum and maintained it. For analysis, information from the first and last second were discarded. This allowed time for the force to reach maximum and the mean force to be measured over three seconds (Chaffin and Andersson, 1991). As soon as the target direction was achieved, feedback on the computer monitor stopped. In order to ensure that the direction of maximal force had not deviated too far during the five-second period, it was continuously recorded. If the mean horizontal or vertical angle from the required direction was 6° or greater over the measured three-second period, the exertion was repeated. The subjects were given 60 - 120 seconds rest between exertions to prevent fatigue (Chaffin and Andersson, 1991).

4.2.3.2 Day 2

The purpose of the second session was to repeat the exertions while emg recordings were being taken from the right lower limb. Once the electrodes had been applied, connected and the emg signals checked, the subject was again presented with the six conditions (Table 4.1) in random order. On this occasion, it was necessary for the subject to achieve the required direction *and* at least 60% of the previous session's maximum force before the audible signal was heard. This ensured that the emg trace was not recorded while the force was of small magnitude. The first second recorded was therefore not discarded, because the magnitude of force was already over 60% of maximum.

Because accurate postural data was available from the Chapter 3 experiment, where posture was observed to depend on the direction of exertion (compare postures of subjects at 1m handle height for exertions to the right (Figure 3.10), forward (Figure 3.14b) and to the left (Figure 3.15a), for example), it was not thought necessary to record accurate postural detail in this experiment. Therefore, the subject's posture during the entire Day 2 session was recorded only on video tape. A bitmap image of the (static) posture of each subject at one instant (when the posture was steady) during the period when the emg was being recorded for each exertion was digitized using the Techno-I video digitizer on an Archimedes A5000 computer. A program was written and used to locate *x* and *y* co-ordinates of the

points shown in Figure 4.1 so that a simple postural stick figure representation was obtained for each subject in each exertion. The points digitized were: head, C7, right and left acromion, right and left elbow, right and left hand, mid-pelvis, right and left knee, right and left ankle, front of right and left foot, back of right and left foot.

4.3 SURFACE ELECTROMYOGRAPHY RESULTS.

4.3.1 Electromyography.

The mean emg value (over four seconds) for each subject and exertion was used in analysis. It was normalized by finding the mean of the values for one muscle group in all exertions for each subject. The individual emg values for that subject were then expressed as a percentage of this mean. The means of these emg percentages with standard error for the right lower limbs of the ten subjects exerting in the six different combinations of direction and hand on the handle are given in Table 4.2.

Table 4.2: Mean and standard error of emg results for ten subjects. F - forward, B - backward, L - to the left, R - to the right, (r) - right hand on handle, (l) - left hand on handle.

MUSCLE		F(r)	B(r)	L(l)	R(r)	R(l)	L(r)
Quads	Mean	93	79	42	193	115	77
	Std Err	10	7	6	20	8	6
Hams	Mean	84	82	139	94	55	146
	Std Err	14	19	20	16	7	14
Tibial	Mean	173	34	37	196	75	86
	Std Err	55	15	12	37	16	21
Gastroc	Mean	50	184	90	127	55	90
	Std Err	15	54	28	27	11	15

Because the variances were large, non-parametric tests were used to detect significant differences between emg records obtained from the four directions. A Wilcoxon signed ranks test was performed to investigate differences between opposing force directions (forward and backward or right and left) and a Friedman test was used to analyze the overall differences between all four directions. The levels of significance are given in Table 4.3.

Table 4.3: Levels of significance for differences between force directions in emg records. F - forward, B - backward, L - left, R - right.

DIRECTIONS	QUADS	HAMS	TIB	GAST
F/B (Wilcoxon)	NS	NS	NS	p<0.01
L/R (Wilcoxon)	p<0.01	p<0.01	p<0.01	NS
ALL (Friedman)	p<0.01	p<0.01	p<0.01	p<0.01

Correlation coefficients were calculated to compare the activity of the four muscle groups. Table 4.4 gives the correlation matrix.

Table 4.4: Correlation matrix showing coefficients and levels of significance between muscle groups for exertions in all directions.

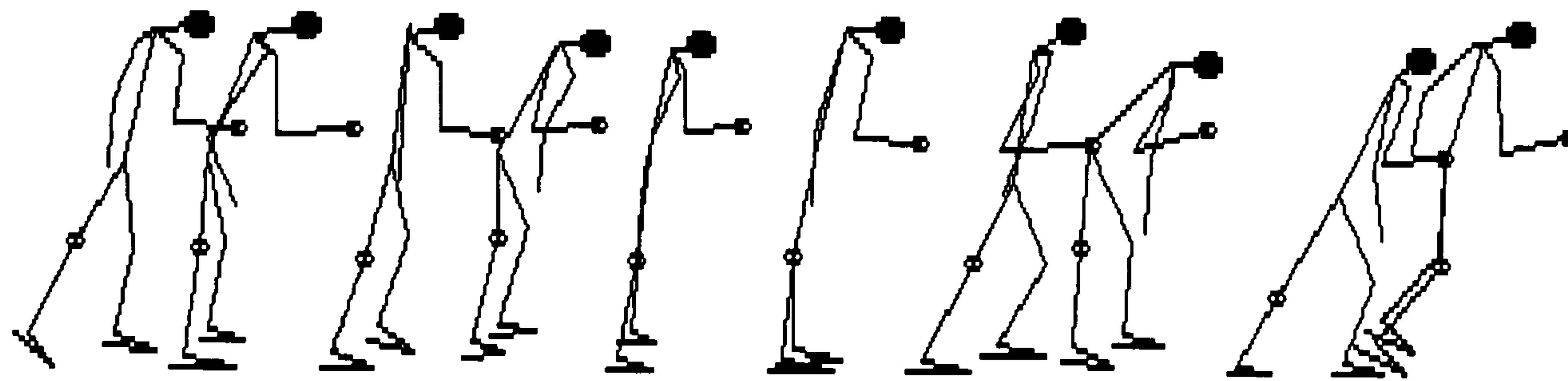
	QUADS	HAMS	TIBIALIS
GASTROC	0.03 (NS)	0.45 (p<0.01)	0.27 (NS)
TIBIALIS	0.60 (p<0.01)	-0.04 (NS)	
HAMS	-0.42 (p<0.01)		

The correlations between the muscle groups generally give results consistent with the expected actions of the muscles. There was a significant negative correlation between quadriceps and hamstrings, which have directly opposing actions at the knee and, in the case of rectus femoris, at the hip. The highest correlation coefficient was found between quadriceps and tibialis anterior. This suggests that in this experiment, extensor activity at the knee and flexor activity at the hip were often accompanied by dorsiflexor activity at the ankle. As expected, the actions of the hamstrings and the gastrocnemius were also significantly correlated, while the pairings of quadriceps and gastrocnemius or hamstrings and tibialis anterior were not. Although tibialis and gastrocnemius would be expected to have opposing actions at the ankle, the relationship between the two muscles was not significantly negatively correlated. However, the actions of gastrocnemius may be complicated by the fact that it is a two-joint muscle and also has actions at the knee.

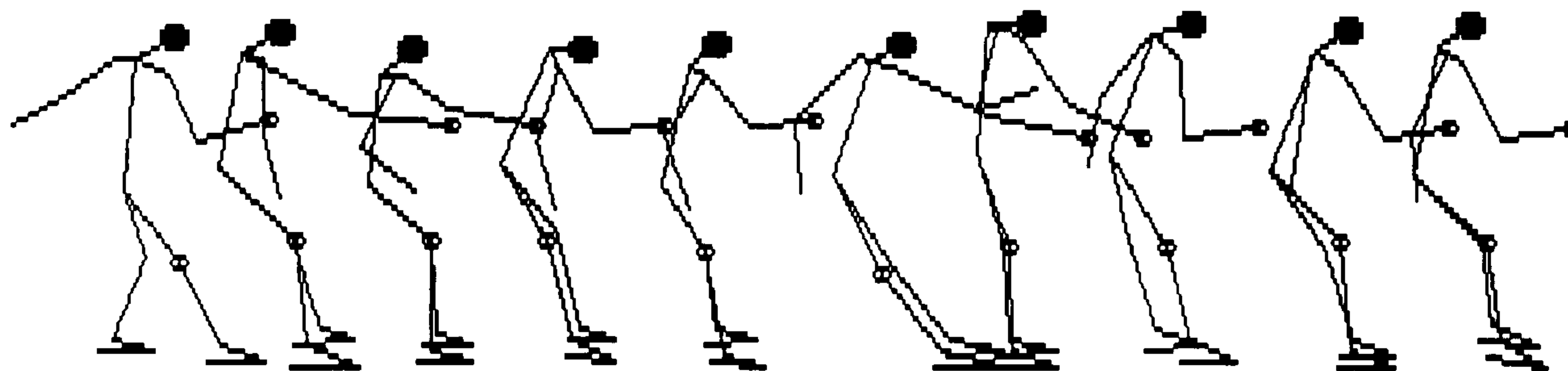
4.3.2 Posture.

Stick figure representations of the postures of all subjects exerting in the six experimental conditions are shown in Figures 4.4 to 4.6.

The use of a biomechanical model to perform a detailed postural analysis was not feasible, since the postures recorded from the emg experiment were simple video records using a camera near to the subject without close control of the optical axis and no provision for parallax errors in measurement. However, accurate postural data from the experiment in Chapter 3 were available. To confirm that the postures of the subjects in both experiments were similar, a mean posture for the eleven subjects in the first experiment and the ten in the second were compared for the four exertions - pushing forward, pulling backward, exerting to the right and to the left. In the emg experiment, the posture while exerting to the right with the right hand was assumed to be a mirror image of that to the left with the left hand. Similarly, it was assumed that an exertion to the right with the left hand was a mirror image of an exertion to the left with the right hand. The mean co-ordinates of the postures while exerting to the left and to the right were therefore calculated from the co-ordinates of 2 x 10 subjects, whereas those for the exertions forward and backward were calculated from ten subjects. Figures 4.7 and 4.8 show the mean postures of both sets of experiments compared in the four directions with the handle at a height of 1m. No statistical correlation of the co-ordinate points is justified, again because the data from the second experiment were taken from simple digitization of a video picture. It is nevertheless clear that although the

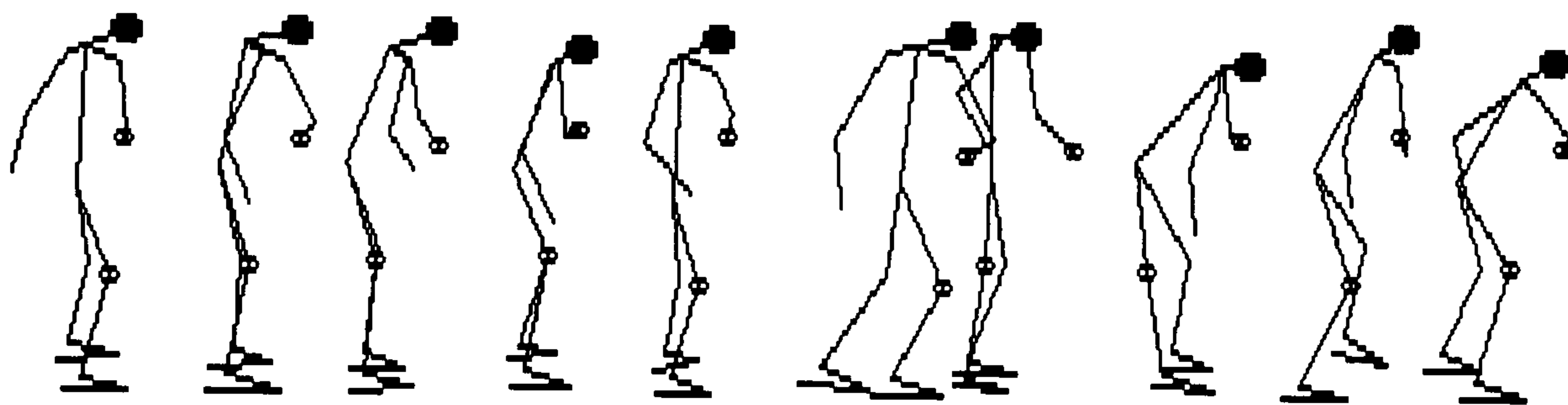


a) Forward

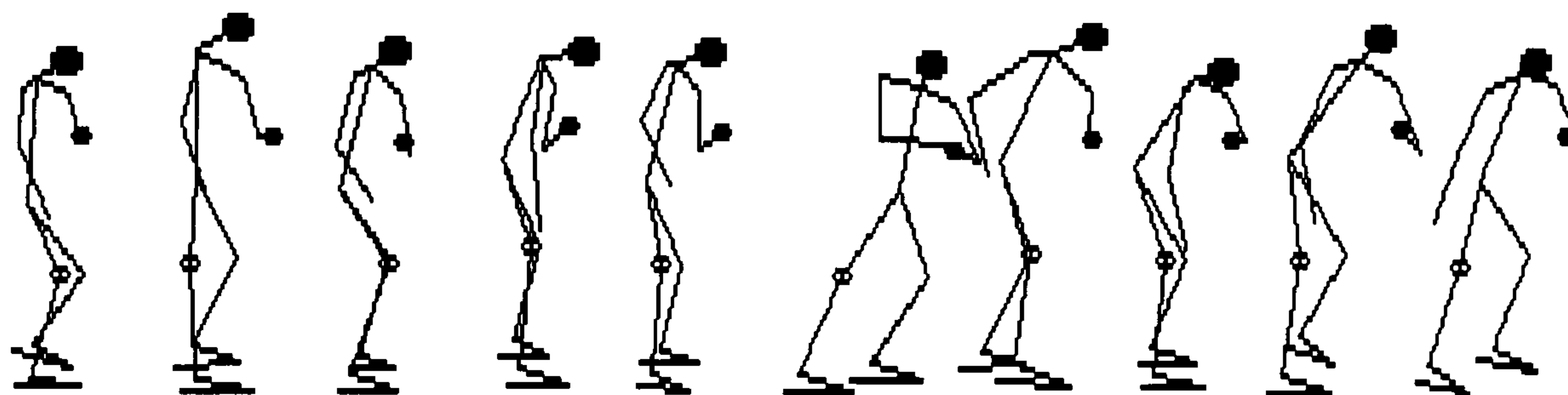


b) Backward

Figure 4.4: Stick figure representations of the postures of all subjects
a) pushing horizontally forward, and b) pulling horizontally
backward. (The right knee is circled.)

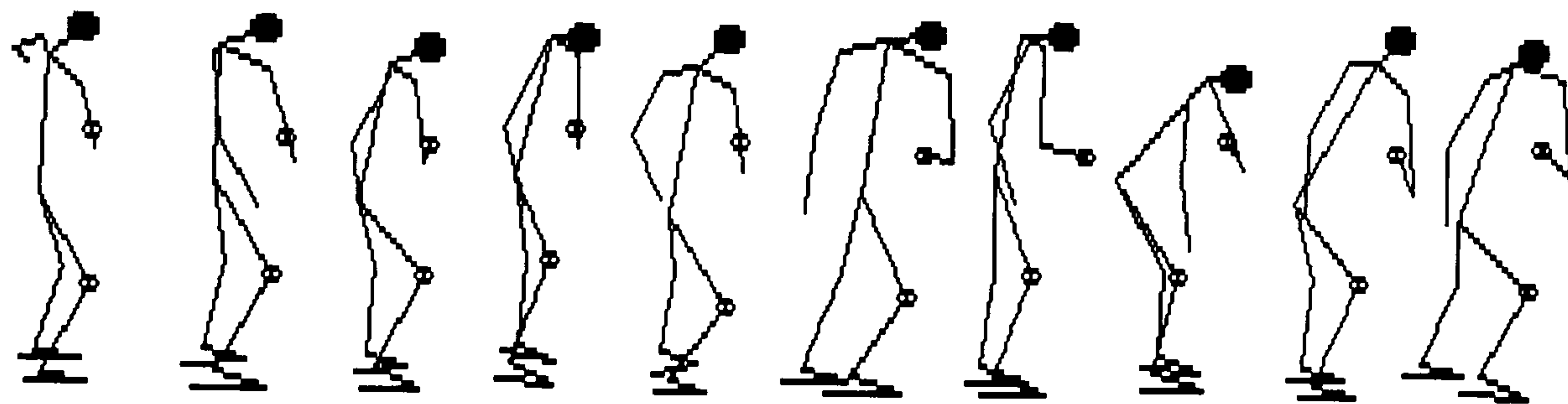


a) To the left (right hand)

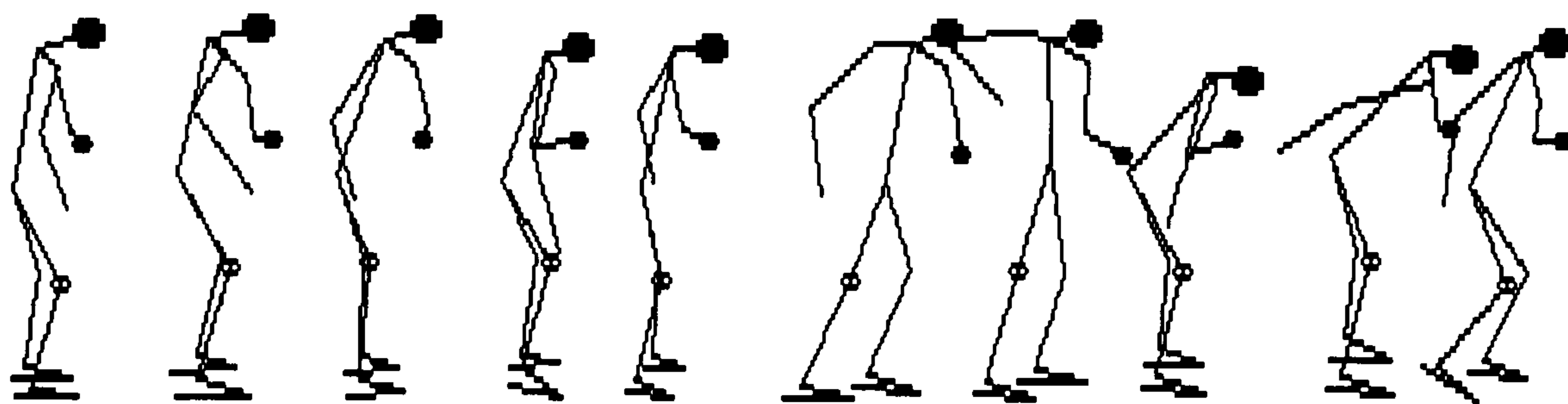


b) To the left (left hand)

Figure 4.5: Stick figure representations of the posture of all subjects exerting to the left with a) the right hand, and b) the left hand. (The right knee is circled.)

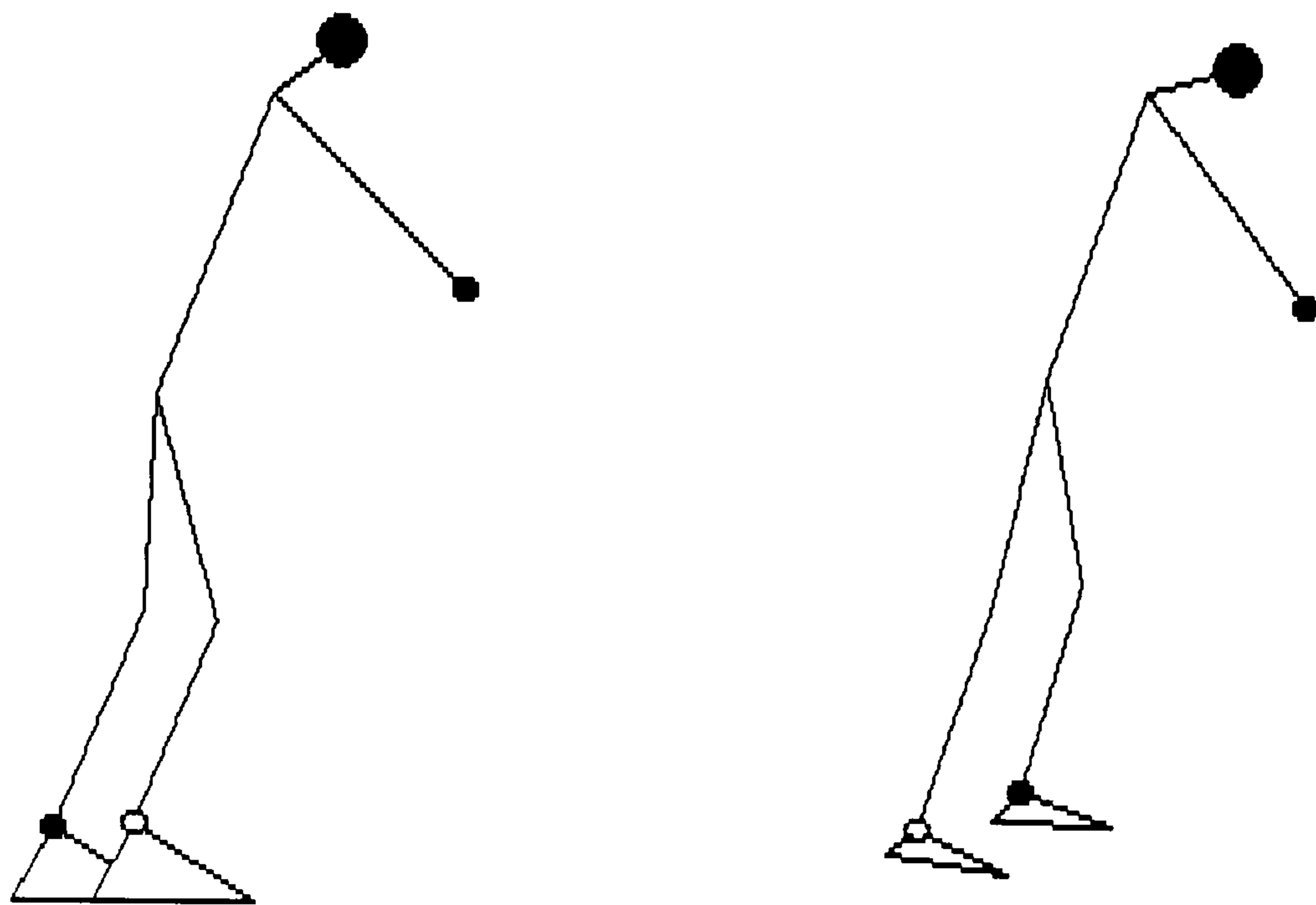


a) To the right (right hand)

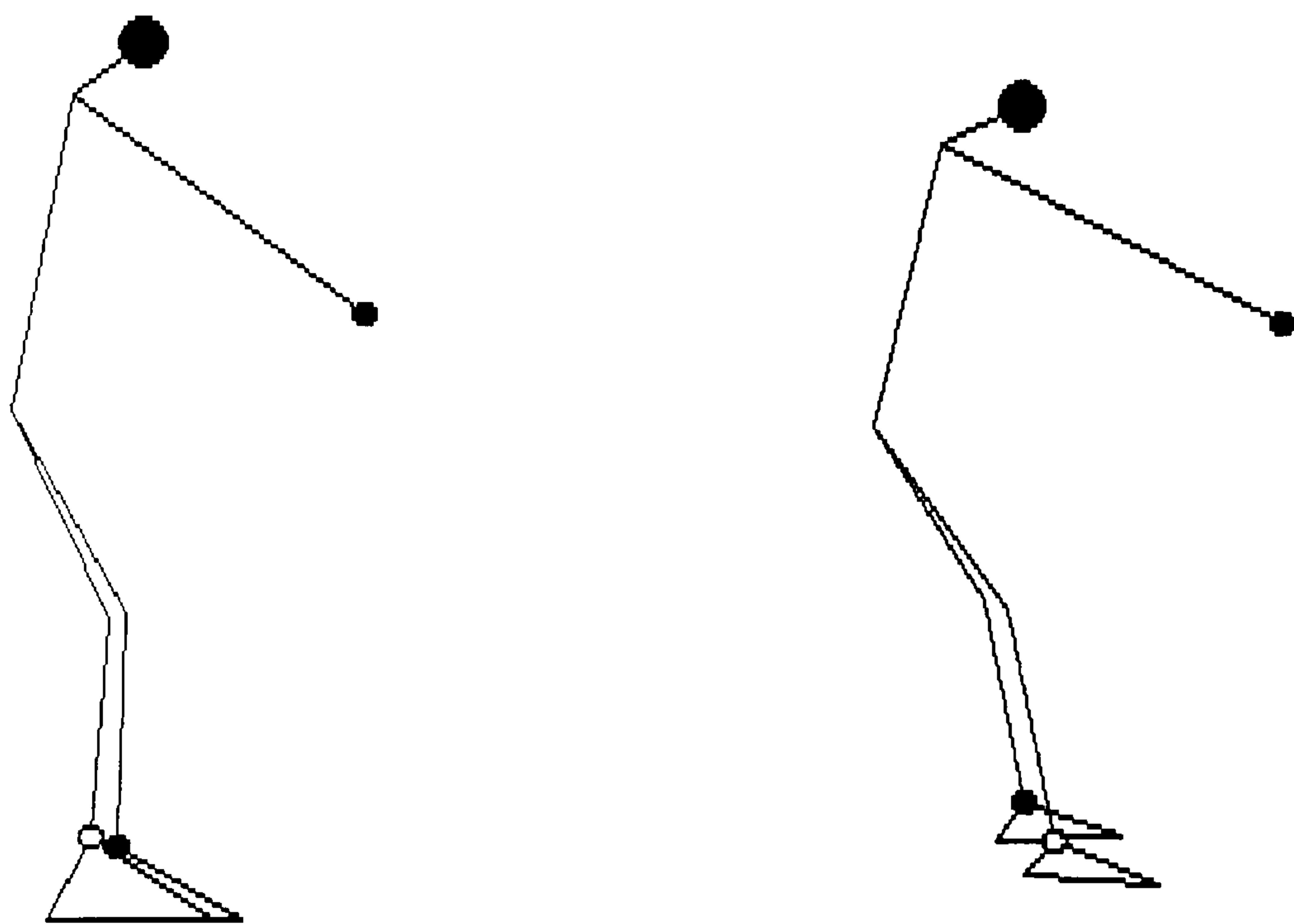


b) To the right (left hand)

Figure 4.6: Stick figure representations of the postures of all subjects exerting to the right with a) the right hand b) the left hand. (The right knee is circled.)



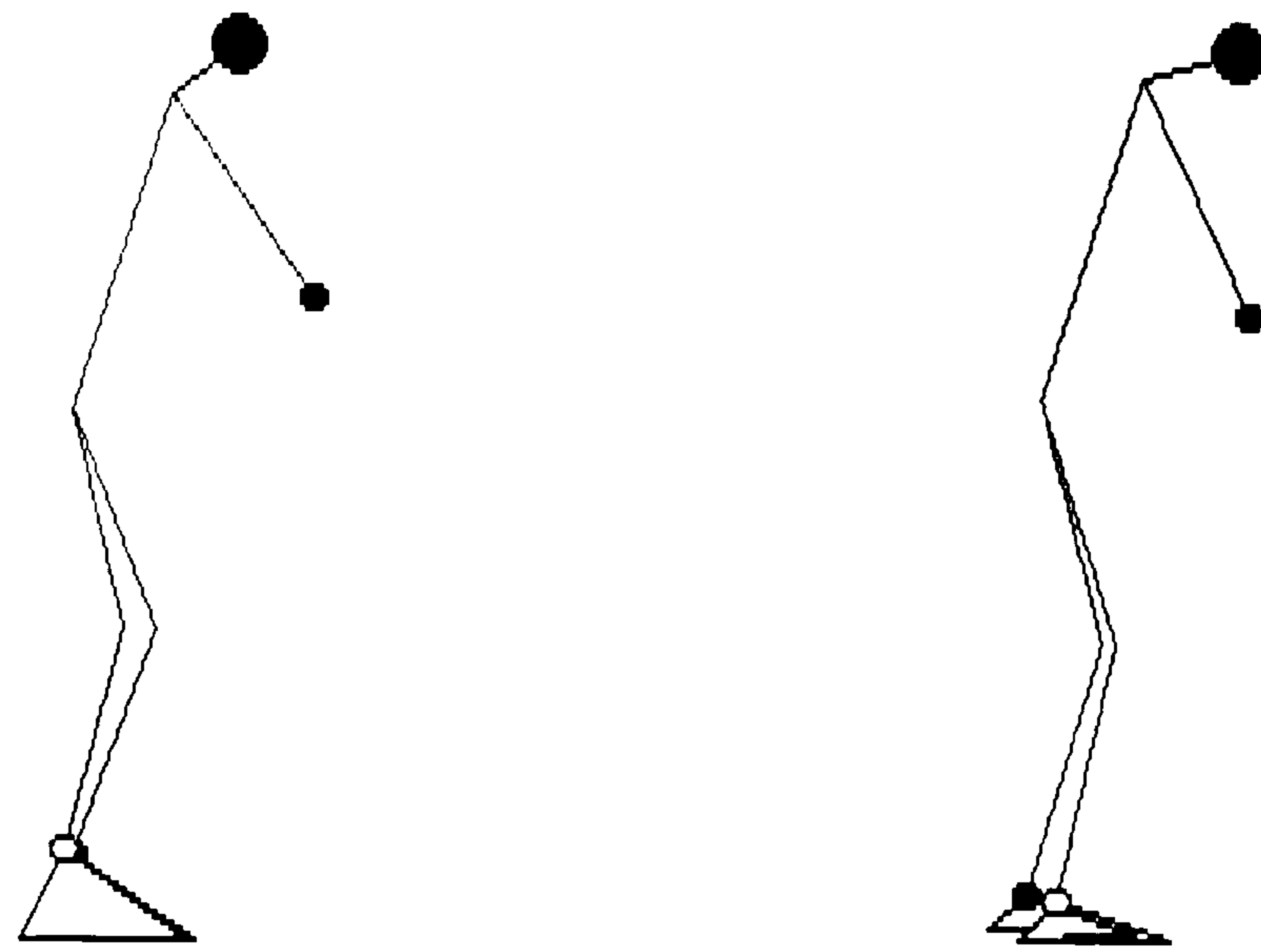
a) Horizontal and Forward



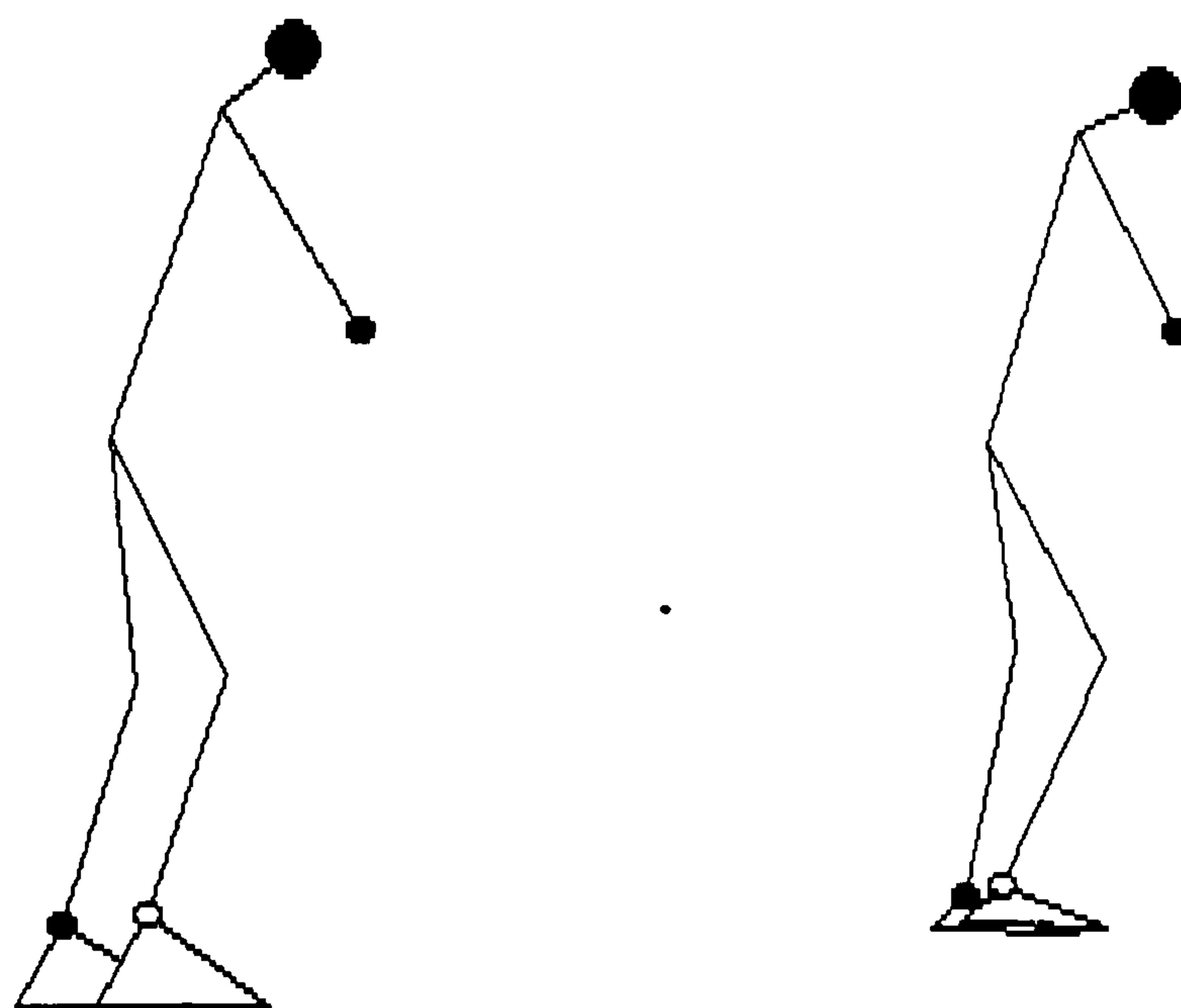
b) Horizontal and Backward

Figure 4.7: The mean postures of subjects exerting a) forward and b) backward.

The stick figures on the left represent the mean postures of the subjects in Chapter 3 and those on the right represent the mean postures of those in the emg experiment.



a) Horizontal and to the Left



b) Horizontal and to the Right

Figure 4.8: The mean postures of subjects exerting a) to the left and b) to the right. The stick figures on the left represent the mean postures of the subjects in Chapter 3 and those on the right represent the mean postures of those in the emg experiment.

posture for each force direction is different, the postures for any one force direction are strikingly similar in both experiments. Therefore, linking postural data from the Chapter 3 experiment to emg data in this study appears justified.

4.3.3 Muscle activity and postural demands.

For simplification, the mean emg readings of the ten subjects were given scores on a scale of 1 to 4 depending on its magnitude. The scaling was as follows:

≤ 70	----	1
71 - 100	----	2
101 - 130	----	3
≥ 131	----	4

It was only possible with the available equipment to record emg signals from one lower limb (the right). It was assumed that the activity of the left lower limb mirrored that of the right when a subject exerted in the opposite direction with the left hand. This allowed emg results for both lower limbs to be presented for exertions to the left and to the right with the right hand (Table 4.5).

The muscle activity measured by the emg was a reflection of the demands due to both posture (weight of body segments) and externally applied forces (reactive force at the hands). To examine these demands, the moments about the pelvis in the fore-aft plane for a subject of mean body weight and applied force (emg

experiment) with a mean posture (Chapter 3 experiment) were calculated. The mean moments are given in Table 4.5 together with the mean emg scores for each muscle and exertion.

Table 4.5: Mean emg scores (expressed on a 4-point scale) and moments about the pelvis in the fore-aft plane. (Positive moments require extensor activity).

DIRECTION	LOWER LIMB	EMG SCORES				MOMENTS (pelvis)
		Quad	Ham	Tib	Gast	
Forward	Right	2	2	4	1	46Nm
Backward	Right	2	2	1	4	208Nm
Right	Left	1	4	1	2	115Nm
Right	Right	4	2	4	3	
Left	Left	3	1	2	1	72Nm
Left	Right	2	4	2	2	

There was no significant difference (see Table 4.3, p.130) in the emg values for the quadriceps or hamstrings between exertions forward and backward (all mean scores = 2), even though a difference might be expected for the hip extensor group, in view of the fact that the moment demands at the pelvis are greater during backward pulls. However, during exertions to the right or the left, the values for the two muscle groups were markedly different. For a subject exerting to the right, Mechanism A would require activity of the flexors of the right hip and extensors of

the right knee together with extensors of the left hip and flexors of the left knee. This implies increased quadriceps activity in the right lower limb and increased hamstring activity in the left. Figure 4.9b shows that the pattern of emg scores in the right thigh was: quadriceps = 4, hamstrings = 2; while in the left thigh it was: quadriceps = 1, hamstrings = 4. For exertion to the left, the reverse conditions would be expected (Figure 4.9a). The readings for the right thigh were then: quadriceps = 2, hamstrings = 4; and for the left thigh were: quadriceps = 3, hamstrings = 1. These results are consistent with the muscle activity expected due to Mechanism A.

The emg scores for tibialis anterior indicate that it is significantly more active (4 and 2) when the foot is pushing forward (right lower limb when exerting to the right and left lower limb when exerting to the left - Figure 4.9) than when it is pushing backward (1 and 2). The scores for gastrocnemius do not show a clear pattern, with scores of 2 and 2 when the foot is pushing backward and 3 and 1 when pushing forward.

During exertions backward and forward, the emg scores for tibialis anterior were 1 and 4 and for gastrocnemius 4 and 1 respectively. Although the difference between the two scores for tibialis were just outside significance level, these results suggest that the two muscles were working oppositely. During pushing, tibialis appeared to be active, which, with the foot fixed, would dorsiflex the leg forward over the ankle, while in pulling, gastrocnemius appeared to be more active.

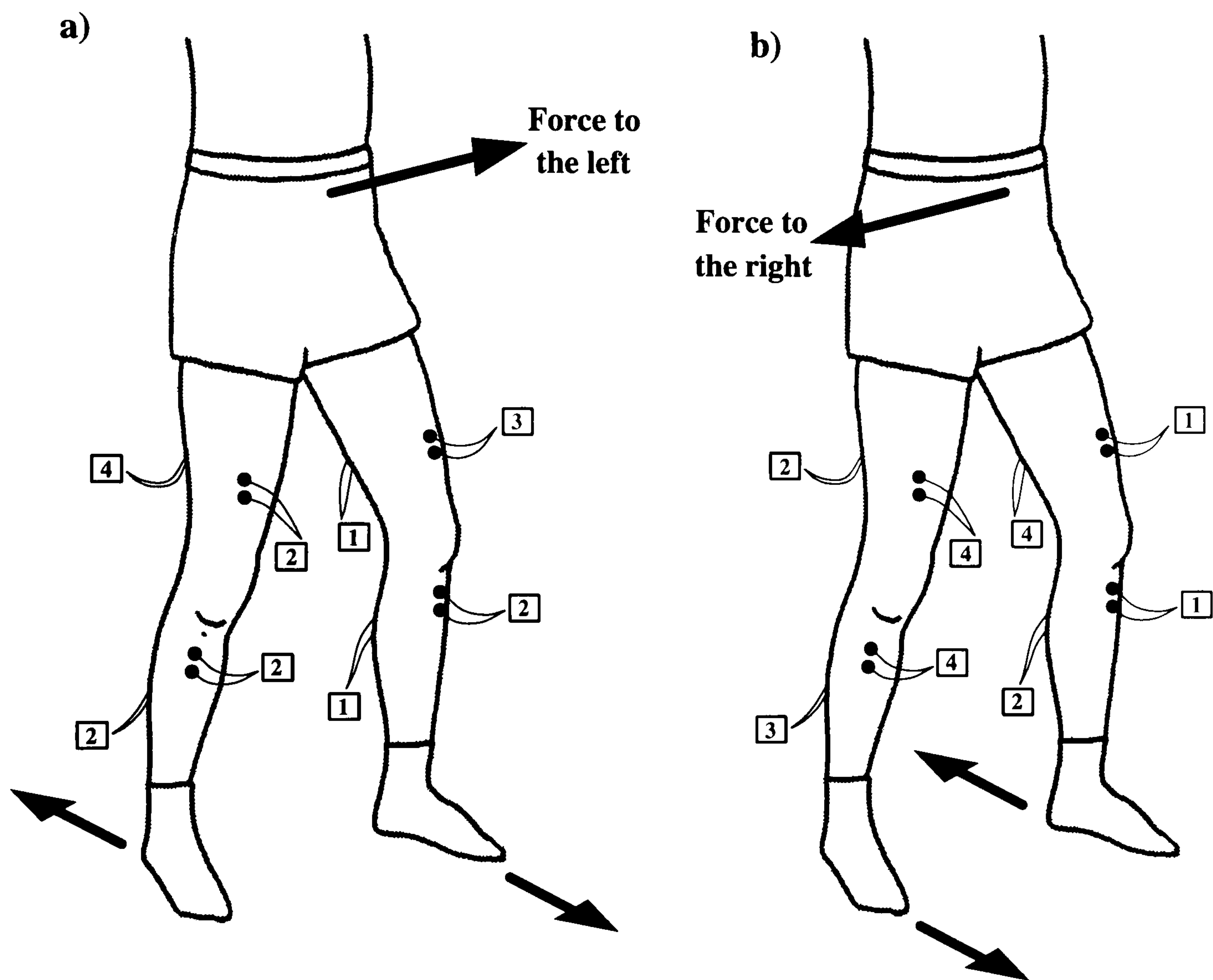


Figure 4.9: Electromyographic scores of muscle groups of lower limbs, showing presumed action of Mechanism A for exertion with the right hand
a) to the left and b) to the right.

The significant differences between all four directions in the four muscle groups confirm that the activities of the muscles differ depending on the direction of exertion.

4.4 SURFACE ELECTROMYOGRAPHY DISCUSSION.

It is realized that a study of the anterior and posterior muscle groups of the thigh and leg cannot clarify all the actions of the lower limb muscles in the generation of a horizontal moment simultaneously with a lateral translational force during exertions to the left and right. Implied in these actions is activity in one lower limb of extensors of the hip with flexors of the knee (Mechanism A), lateral rotators (Mechanism B) and abductors, and in the other, flexors of the hip with extensors of the knee, medial rotators and adductors. The inaccessibility of the lateral rotators prohibited their investigation and the role of the adductors and abductors was not investigated. Nevertheless, the use of surface emg to investigate the activity of quadriceps, the hamstrings, tibialis anterior and gastrocnemius has supported the suggestion that Mechanism A is partly responsible for the generation of the moment at the feet. Rectus femoris and the hamstrings are two-joint muscles, which have actions at both the hip and knee. The activity of these muscles was increased substantially when their action would have been required for Mechanism A, despite the fact that (at least for the hamstrings) the postural demands varied considerably. It is not possible to speculate on the potential activity of the lateral and medial hamstrings to rotate the leg laterally or medially, because the muscles were investigated as a group.

It is difficult to predict the actions at the ankle when one foot is exerting a backward or forward force upon the floor. As discussed previously, the muscles crossing the joint may simply stabilize it while the necessary force is transmitted from muscle activity higher up in the leg and thigh. It is also possible that muscles about the ankle have a more active role in assisting the production of the forward or backward force. The emg results from the gastrocnemius during force exertions to the left or right are not decisive and calculation of the moments about the knee or ankle joints is not possible since the forces transmitted through each lower limb separately cannot be determined during such asymmetrical activities. The actions of this muscle are complicated by the fact that it crosses two joints, and measured emg activity may reflect requirements at the knee, ankle or both. The significantly positive correlation of gastrocnemius with the hamstrings suggests that both muscles acted together during flexion of the knee. Tibialis anterior was more active when the presumed exerted force on the floor was forward and its emg results correlate significantly with those of quadriceps.

For forces at the hands directed forward or backward, the mean emg scores indicated that the usual strategy for pushing forward was tibialis anterior dorsiflexing the leg forward over the foot, while the gastrocnemius was employed for propulsion during backward pulls.

Some of the variance between emg results can be explained by the postural strategies that subjects employed to exert forces. For example, subjects who pushed forward with a plantarflexed ankle had greater emg readings for

gastrocnemius than tibialis anterior, while those who raised the anterior part of the foot off the ground had higher readings for tibialis anterior for the same exertion.

In this study, the postures of the emg subject group were assumed to have the same posture as the subject group from Chapter 3. A comparison of the mean posture of the two subject groups appears to justify this assumption.

4.4.1 Surface electromyography.

Various studies have shown that altering the length of a muscle affects the magnitude of the emg signal (see section 2.7.3.1, p.64). For the forces directed laterally, which were of particular interest in this study, the degree of flexion at the hip, knee and ankle was similar in many subjects. Observed differences were mostly in the degree of lean to the right or left, so that the leg was angled away from the vertical in a coronal plane. Under these circumstances, the lengths of the anterior and posterior muscles would not change significantly, except perhaps that of tibialis anterior during pronation and supination of the leg on the foot.

However, the emg score for the right muscle in the pronated (and therefore longer) position during exertions to the right was greater than that when the leg was vertical. The score for the muscle during exertion to the left when the leg was vertical was higher than that when the leg was angled medially (supinated and shorter), despite the fact that a lower reading might be anticipated on account of the muscle being stretched (Inman *et al.* 1952; Grieve and Pheasant, 1976). A higher reading might be expected (and was obtained) when the leg was pronated

over the foot, because the lateral lean of the leg would place a greater postural demand on the muscle.

Another problem that investigators have described is variable emg results depending on the position of the electrodes (Zuniga *et al.* 1970). Every care was taken to ensure that the electrodes in each subject were placed in a similar position. However, the emg readings for a muscle were normalized by comparing recordings from a subject to his own mean, rather than that of other subjects, and all emg readings from one muscle were obtained with the electrodes in the same position.

Cross talk may also cause difficulties in the analysis of surface emg data (Zuniga *et al.* 1970; De Luca and Merletti, 1988). In these experiments, no attempt was made to distinguish emg recordings of muscles lying close together (for example, the individual muscles of the quadriceps). Rather, muscle groups with similar functions were examined together.

4.5 FORCE PLATE - MATERIALS AND METHODS.

4.5.1 Experimental equipment.

The force plate used (Model OR6-5-1 of an Advanced Mechanical Technology Incorporated (AMTI) Computerized Biomechanics Laboratory System) was calibrated before the experiment began. Vertical loads and horizontal forces and moments were applied individually and in combination to the platform in the ranges expected during the experiment. The mean r^2 values for applied forces, moments and positions of the centre of foot pressure were all above 99.9%, while the mean r^2 value for combined forces and moments was 99.6%.

The apparatus (Figure 4.10) consisted of the force plate, a platform, and three pulleys mounted on scaffolding. This arrangement allowed readings to be taken from the force plate while subjects stood with one or both feet on the plate, exerting forces on a handle in different directions.

The floor of the experimental area was divided into left and right halves by a central longitudinal line. The force plate, of dimensions 0.51 x 0.46m, was used under three conditions of foot placement. When both feet were required to be on the force plate, the central longitudinal axis of the plate was positioned over the central longitudinal line of the floor. When either the left or right foot was required to be on the force plate, the plate was moved laterally so that either its left or right edge was over the central line, and a platform of the same height was

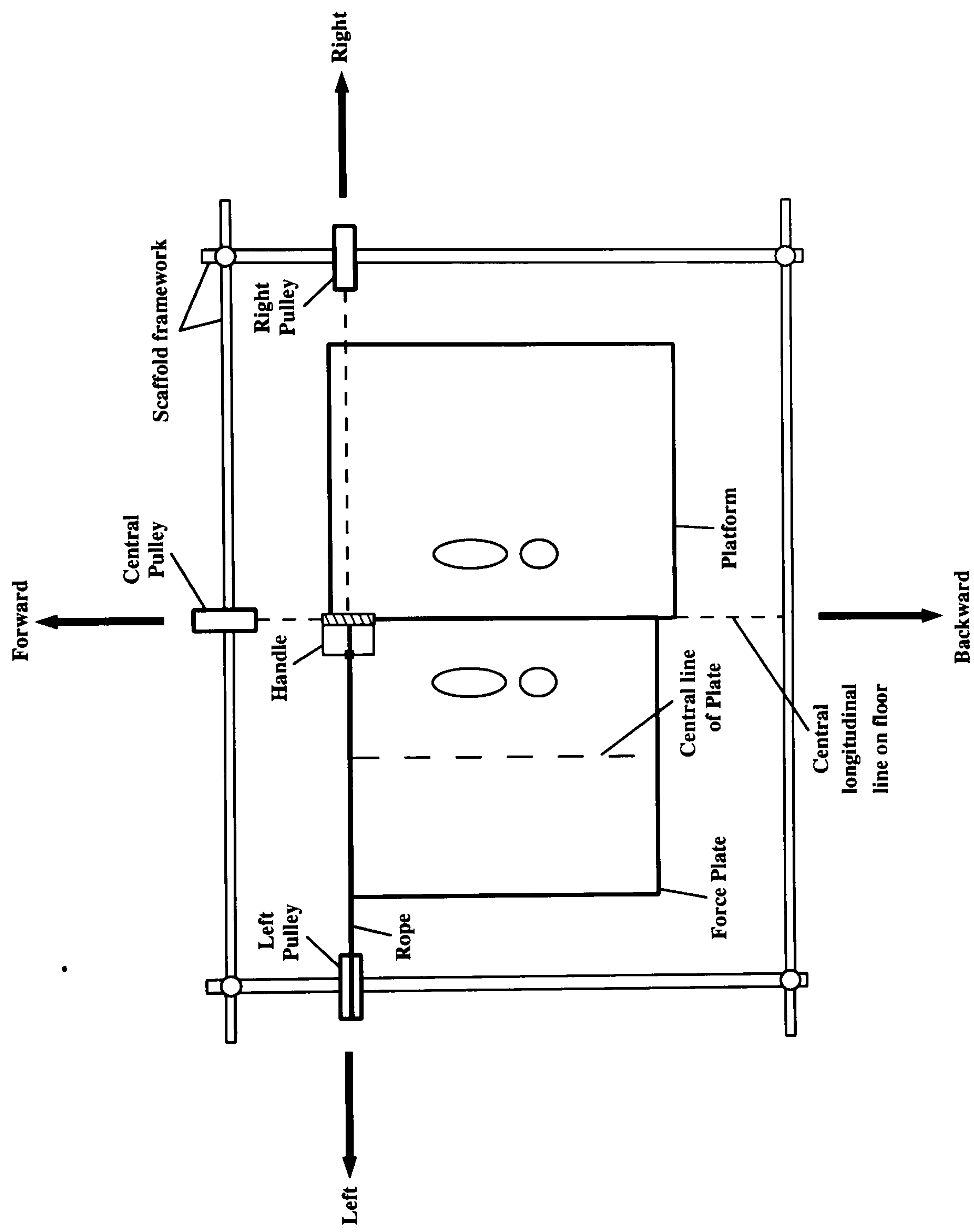


Figure 4.10: Experimental set-up for measuring forces and moments produced by the left foot (viewed from above).

placed immediately adjacent to it on the other side of the line. The platform was prevented from slipping horizontally by bolts which passed through it into holes drilled into the floor. Both the force plate and platform were covered with emery cloth.

Three pulleys were mounted on a scaffold framework in the positions shown in Figure 4.10. The central pulley was in line with the central longitudinal line of the floor and the left and right pulleys were in line with the front edge of the force plate. A weight was suspended on one end of a rope, which then passed over one of the pulleys. A handle of diameter 30mm was mounted on the other end of the rope. This arrangement allowed the subject to exert forces to the right (left pulley), backward (central pulley) and to the left (right pulley) by pulling on the handle. In order for the subject to exert a force in a forward direction, the handle was attached to a 20mm diameter aluminium rod of length 0.79m, which passed through a greased nylon bush mounted immediately above the central pulley. Its other end was attached to a rope, which then passed backward over the pulley and attached to a suspended weight. The top of each pulley (and therefore the horizontal section of the rope and the centre of the handle when in use) was at a height of 1m above the surface of the platform and force plate. Figure 4.10 shows the lay-out when a subject was standing with the left foot on the force plate and exerting to the right.

4.5.2 Subjects.

Fourteen male subjects took part in the study. Their mean age, height and weight were 28.3 years (s.d. 5.2), 1.76m (s.d. 0.08) and 75.6kg (s.d. 14.0).

4.5.3 Experimental procedure.

The subject's age, height and weight were recorded and the procedure was explained. A consent form was signed. For each exertion, the subject was instructed to have each foot with its longitudinal axis parallel to and equidistant from the central longitudinal line on the floor to ensure that the directions forward, backward, right and left could be maintained when describing the direction of exertion. The feet did not necessarily have to be opposite each other, but were not allowed to pass forward of the front of the platform.

Baseline readings were taken from the force plate. The subject then stepped on to the platform and plate and carried out 12 exertions in one of four directions (forward, backward, right or left) with one of three foot configurations (left foot, right foot or both feet on the force plate). Because the plate and platform were difficult to move, the subject completed all four exertions for one foot configuration before moving on to the next. The order of the four directions of exertion varied between each foot configuration and each subject. A period of at least 60 seconds elapsed between each exertion.

When the reading from the force plate was taken, the subject’s hand was in line with all three pulleys (i.e., immediately above the central longitudinal line of the floor and the front of the platform in the same horizontal plane as the pulleys). Either the rope (for exertions to the right, left or backward) or aluminium bar (for exertions forward) had to be horizontal and in line with the corresponding pulley. The orientation of the handle was not fixed in the vertical plane, but was always at a right-angle to the rope or bar. The weight suspended from the end of the rope was supported above the level of the force plate and platform. When the subject performed the exertion, it was necessary for the handle to be moved only a small distance before it reached the required position.

The weight lifted by each subject was calculated from the mean force achieved (as a percentage body weight) during a previous investigation of whole body strength in eleven subjects using the TAFS handle (Pinder *et al.* 1995). For the four directions of exertion in the present study, the mean percentage body weight force from the data of Pinder *et al.* was rounded down to the nearest 5%. Thus the percentage body weight requirement for each of the four directions was:

Forward	-	30%
Backward	-	40%
Left	-	25%
Right	-	35%.

As soon as the subject was stable, with correct hand and foot position, data from the force plate were recorded for 0.2s. The measurements taken (Figure 4.11) were:

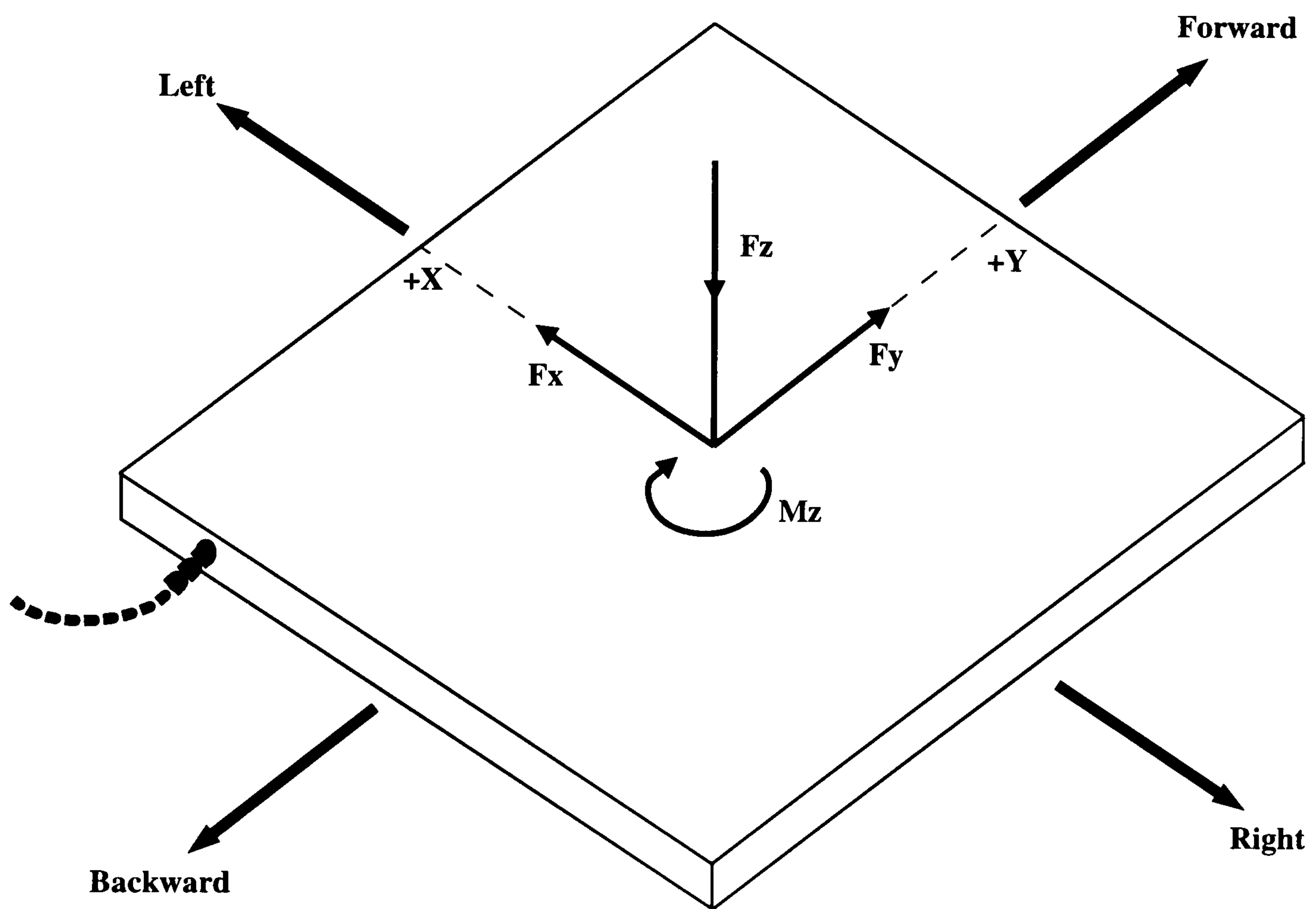


Figure 4.11: Measurements recorded by the force plate: F_x - force in the left-right axis, F_y - force in the forward-backward axis, F_z - force in the up-down axis, M_z - turning moment in the horizontal plane.

- a) horizontal forces in the left-right (F_x) and forward-backward (F_y) axes and vertical forces in the up-down (F_z) axis,
- b) the position of the centre of foot pressure (X, Y) - where (0,0) were the x and y co-ordinates of the centre of the force plate, and
- c) the turning moment in the horizontal plane (M_z).

note: If T_z = pure torque in the horizontal plane, then:

$$M_z = F_y.X - F_x.Y + T_z$$

From this equation, T_z could be calculated.

To test repeatability, one subject performed the experiment twice on separate days. The forces, horizontal moments and positions of the centre of foot pressure measured from the force plate during both trials were compared.

4.6 FORCE PLATE RESULTS.

For each direction of exertion, three recordings were taken - with the left foot only, the right foot only, or both feet in contact with the force plate. In order to compare these three situations, it was assumed that the action of the feet upon the plate/platform was similar for all three conditions. To check that the assumption was reasonable, the readings taken when only the left or right foot was in contact with the force plate were added together (LR), and compared to the readings when both feet were in contact (B). Measurements compared were the forces in the x -, y - and z -axes (F_x , F_y and F_z respectively), the total moment (M_z) and pure torque

(Tz) in the horizontal plane, and the position of the centre of foot pressure in the x- and y-axes (X and Y respectively). The correlation coefficients and levels of significance of the variables compared are given in Table 4.6. In all cases, the correlations were statistically significant. Figures 4.12 to 4.14 show the scatter plots.

Table 4.6: Correlation coefficients (r) and levels of significance (p) between measurements taken with both feet on the force plate (B) and individual measurements of the left and right foot added together (LR). Fx = force in x-axis, Fy = force in y-axis, Fz = force in z-axis, Mz = moment in horizontal plane, Tz = pure torque in horizontal plane, X,Y = co-ordinates of centre of foot pressure in x- and y-axes respectively.

	r	p
FxLR vs FxB	0.96	<0.0001
FyLR vs FyB	0.99	<0.0001
FzLR vs FzB	0.84	<0.0001
MzLR vs MzB	0.89	<0.0001
TzLR vs TzB	0.86	<0.0001
XLR vs XB	0.88	<0.0001
YLR vs YB	0.91	<0.0001

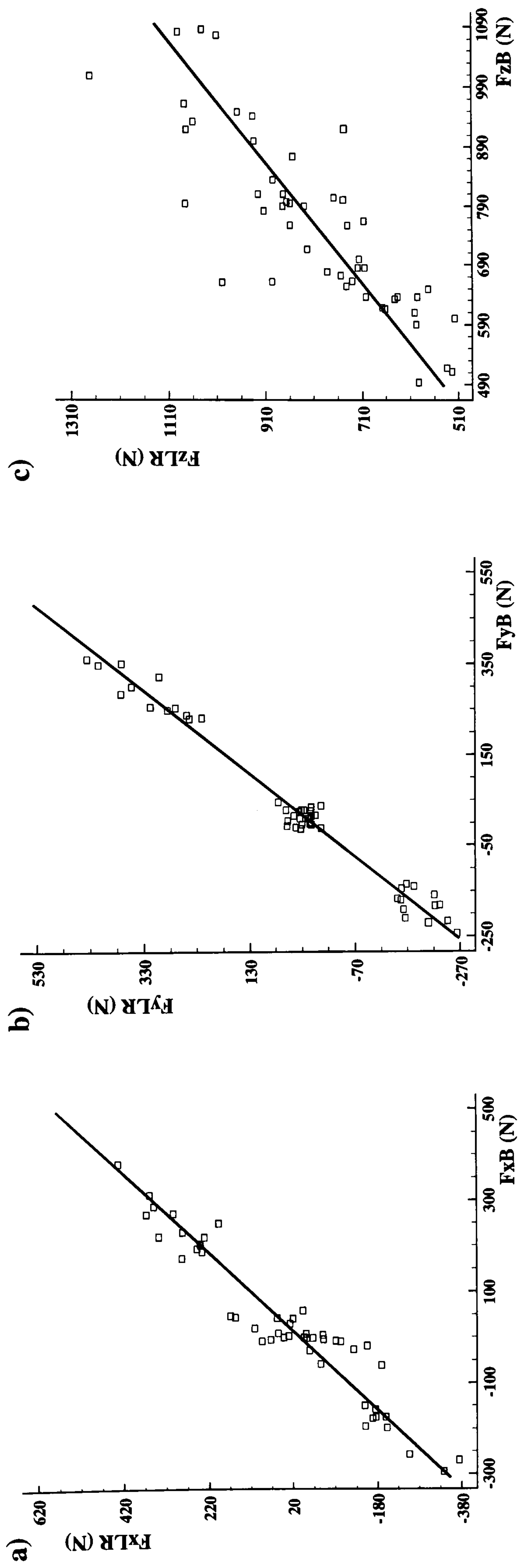


Figure 4.12: Regression of forces of left and right foot added (LR) vs both feet (B) in a) x-axis, b) y-axis and c) z-axis.

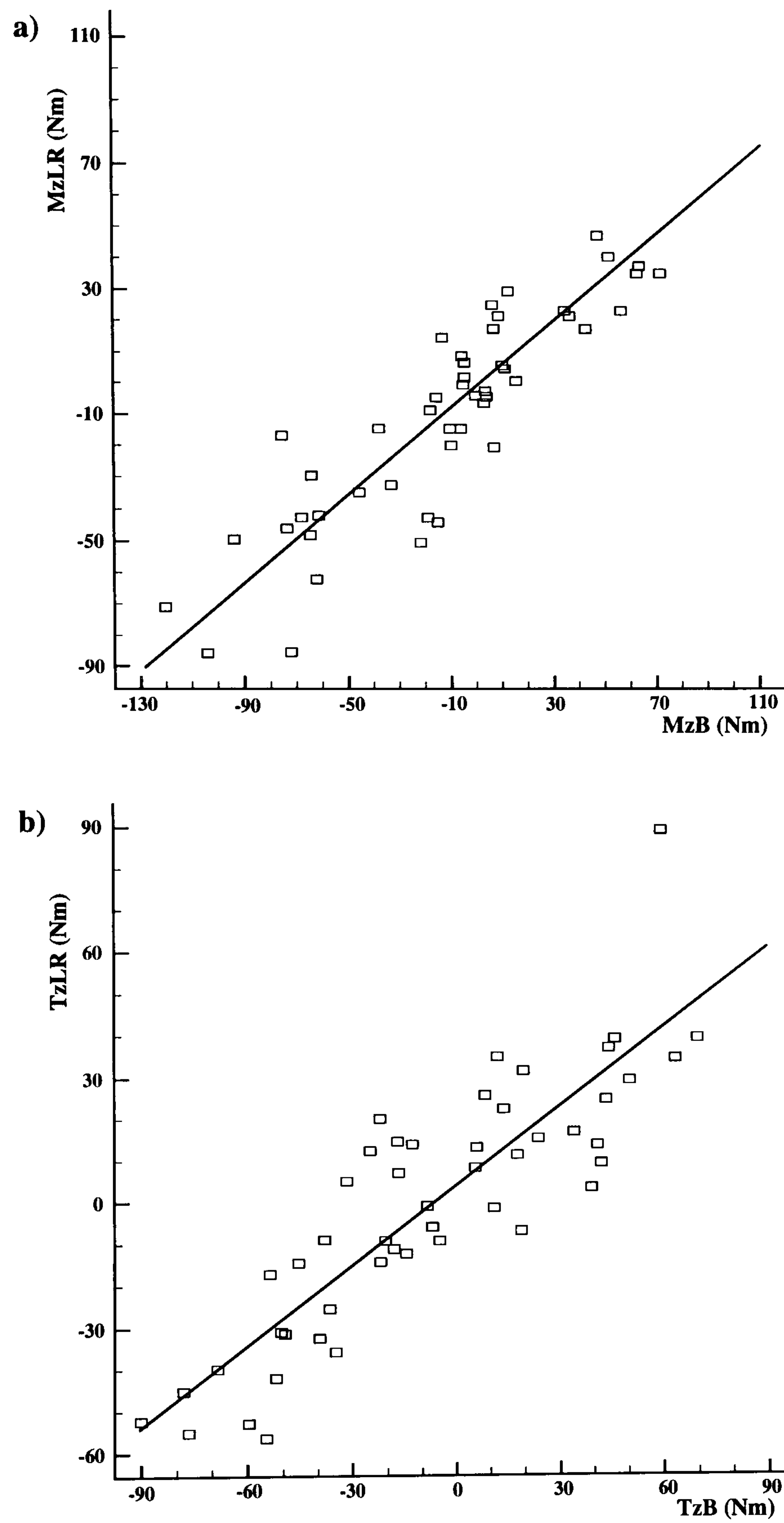


Figure 4.13: Regression of a) total torque (Mz) and b) pure torque (Tz) of left and right foot added (LR) vs both feet together (B).

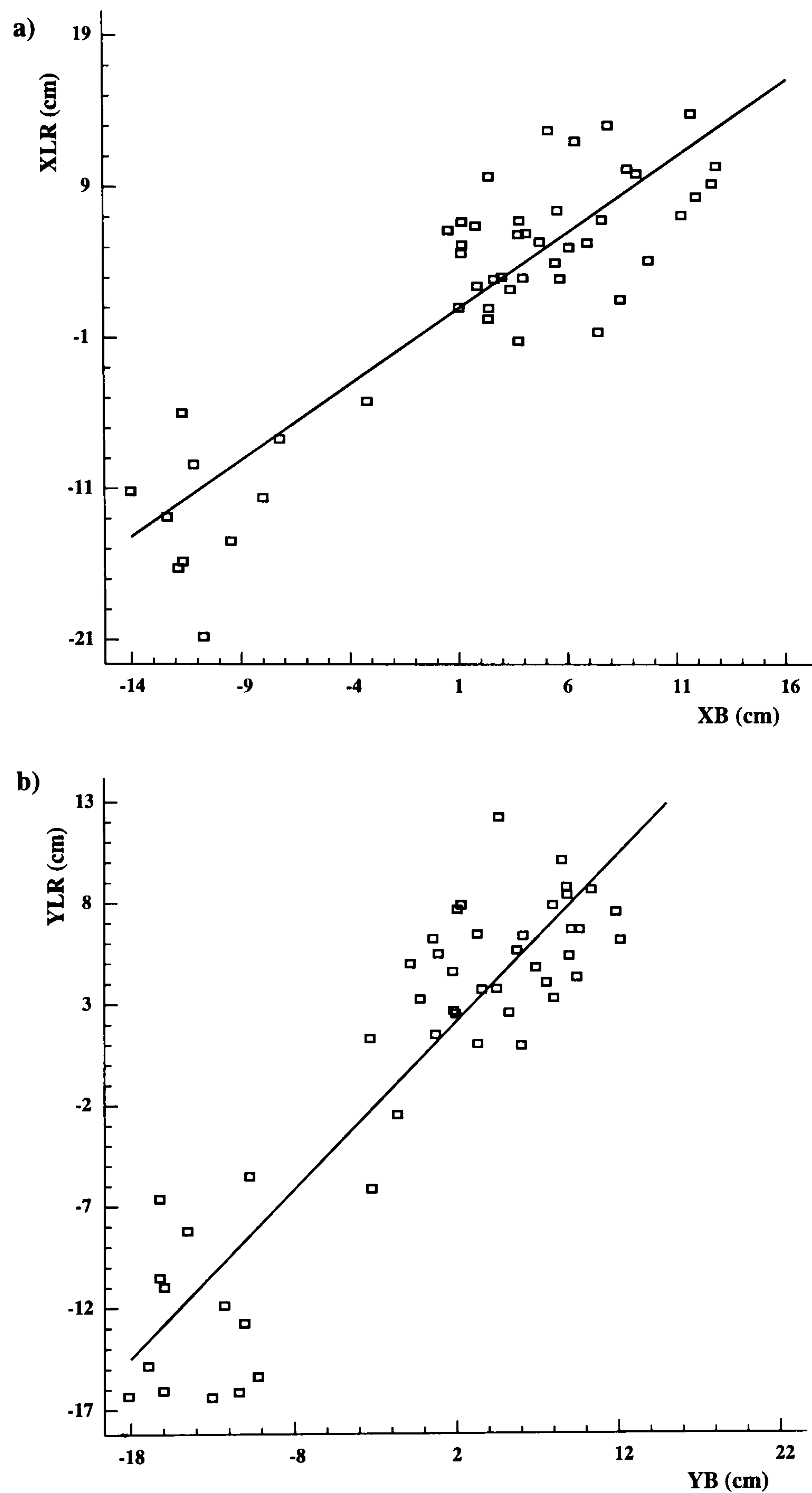


Figure 4.14: Regression of distance of centre of foot pressure from origin of left and right foot added (LR) vs both feet together (B) in a) x-axis and b) y-axis.

The forces at the hands in the x-axis (left-right) or y-axis (forward-backward) during exertions were known, and the translational forces at the feet were measured by the force plate. The horizontal moments expected at the feet during left and right exertions could be calculated from the product of the force at the hand and the distance between the hand and the centre of foot pressure. The actual moments were measured by the force plate. A comparison could thus be made of a) the forces at the hands in the x- and y-axes against those measured at the feet, and b) the expected moments at the feet against those actually measured. The correlation coefficients and levels of significance are tabulated in Table 4.7 and the scatter plots are shown in Figure 4.15. The scatter seen in Figure 4.15a is a result of the subjects applying forces that are fixed percentages of body weight, rather than fixed weights.

Table 4.7: Correlation coefficients (r) and levels of significance (p) between forces at the hand and at the feet, and between expected moments at the feet and those actually measured.

	r	p
Forces	0.99	<0.0001
Moments	0.93	<0.0001

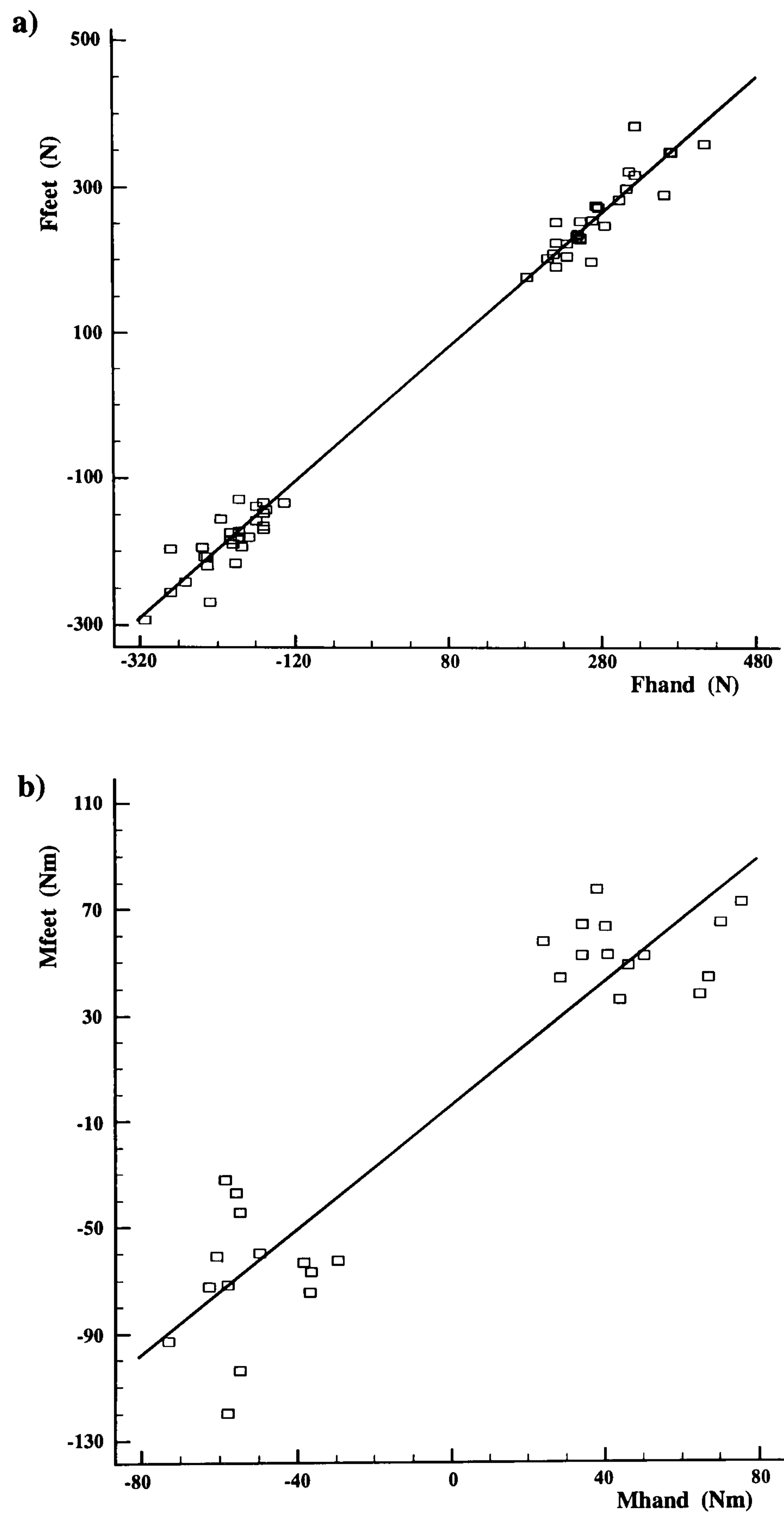


Figure 4.15: Regression of a) force at hands vs force at feet, and b) expected vs measured horizontal moment at feet.

As discussed previously, the total horizontal moment at the feet may be produced by a force at each foot acting in a direction opposite to the other (Mechanism A) or both feet exerting an individual torque on the ground (Mechanism B). These factors could be separated when only one foot was on the force plate, and, if added together, should be similar to the total moment measured when both feet were on the plate. The correlation coefficient for this comparison was 0.92 ($p<0.0001$). The scatter plot and regression line are shown in Figure 4.16. This again supports the view that adding the measurements made when the left or right foot was in contact with the force plate is similar to the conditions present when both feet were in contact.

Also, the factors making up the total moment can be isolated, and their percentage contribution to the total moment calculated (Table 4.8).

Table 4.8: Actual and percentage contribution of factors to total horizontal moment. Mech A = Mechanism A, Tz(L) = pure torque exerted by left foot, Tz(R) = pure torque exerted by right foot.

Contrib	Direction	Mech A	Tz(L)	Tz(R)	Total
Actual (Nm)	RIGHT	-33	-26	-11	-70
	LEFT	36	11	22	69
Percent	RIGHT	47	37	16	100
	LEFT	52	16	32	100

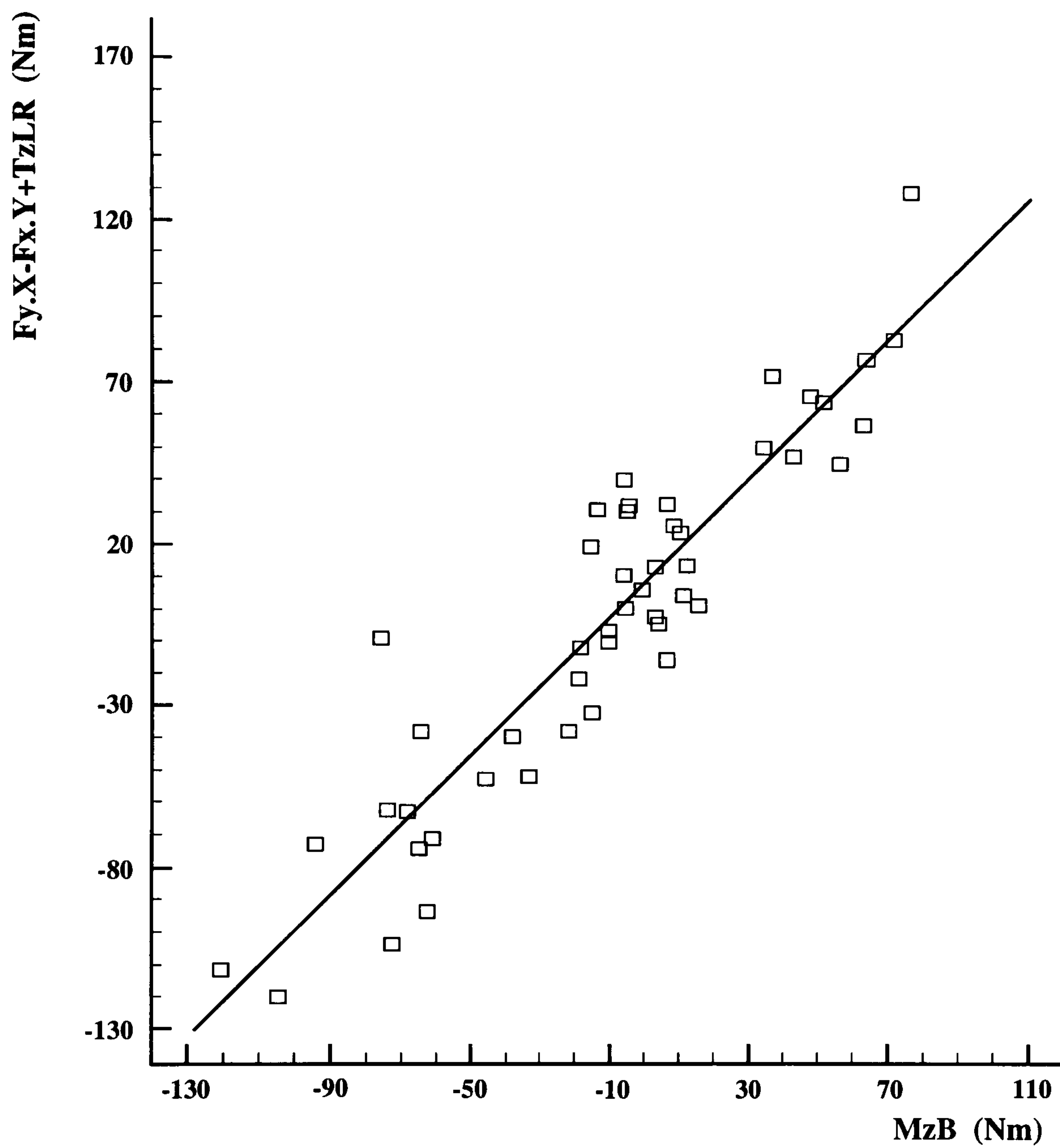
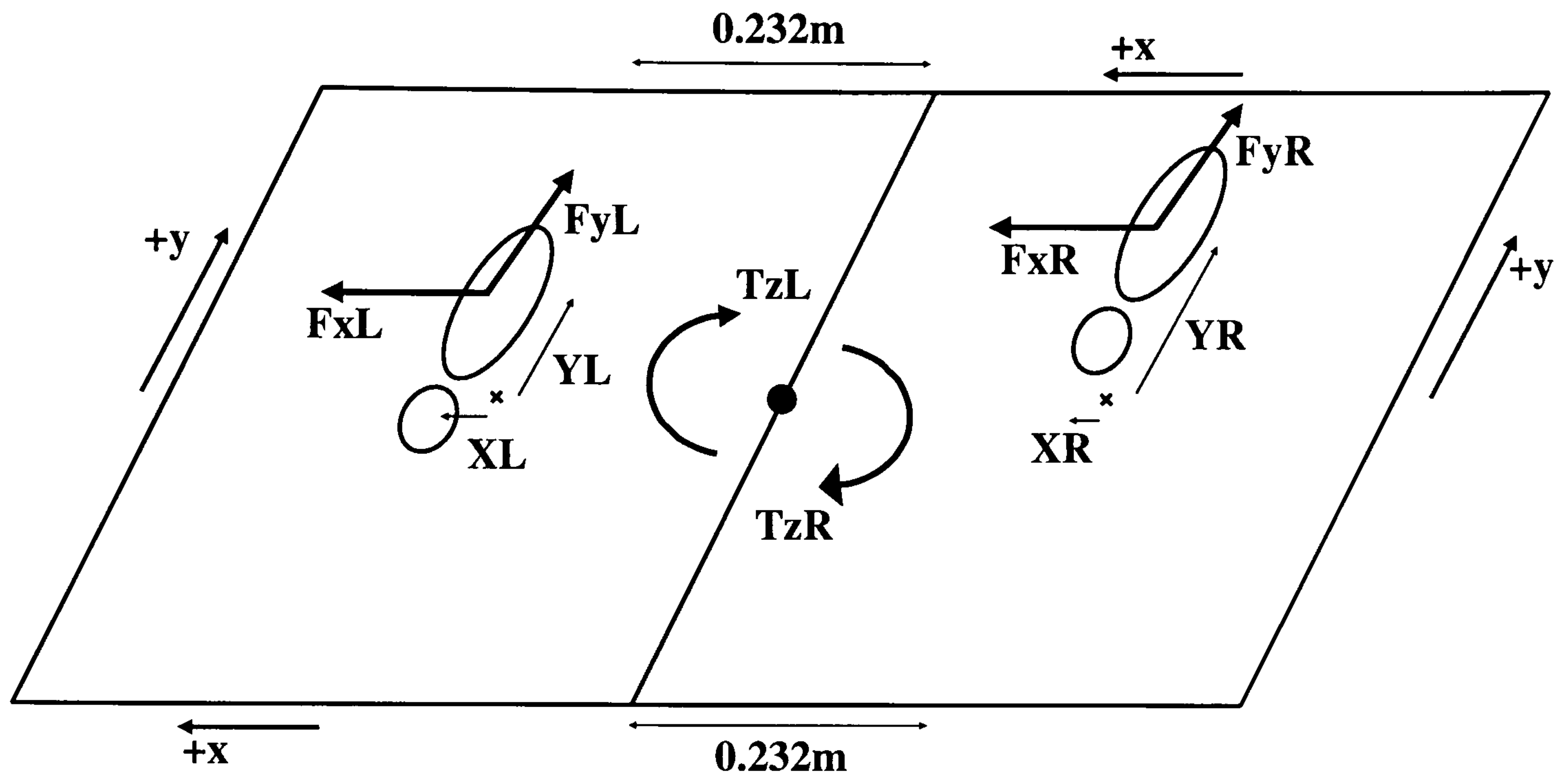


Figure 4.16: Regression of the sum of the contribution of individual factors to the total horizontal moment vs the total moment (Mz) measured by the force plate with both feet on the plate.



The figure depicts the conditions when the left and right feet were on the force plate separately, but as though the force plates had been placed adjacent to each other. The contribution of factors making up the total horizontal moment at the feet could be calculated by taking moments about the hypothetical centre of the plate (●), so that:

$$F_{yL}(0.232 + XL) - F_{xL} \cdot YL - F_{yR}(0.232 - XR) - F_{xR} \cdot YR + T_{zL} + T_{zR}$$

where L - left foot, R - right foot, F_x and X - forces and distances in the left-right axis, F_y and Y - forces and distances in the forward-backward axis, and T_z - pure moment in the horizontal plane.

The sum of F_{yL} , F_{xL} , F_{yR} and F_{xR} represents Mechanism A, while T_{zL} and T_{zR} represent Mechanism B. The contribution of these factors in Nm and as a percentage of the total moment could then be calculated (Table 4.8).

These results show that Mechanism A accounts for approximately 50% of the horizontal torque generated during an exertion to the right or left. The remaining 50% was generated by Mechanism B, approximately two thirds of which was exerted by the right lower limb when pushing to the left or the left lower limb when pushing to the right and one third by the right lower limb when pushing to the right or the left lower limb when pushing to the left.

4.6.1 Repeatability.

The correlation coefficients between measurements from the force plate in the first and second trials of one subject were 0.98 ($p < 0.001$) for the forces, 0.83 ($p < 0.001$) for the horizontal moments and 0.64 ($p < 0.001$) for the x and y co-ordinates of the positions of the centre of foot pressure.

4.7 FORCE PLATE DISCUSSION.

The results of this study show that both Mechanisms A and B provide the method for generation of horizontal torque at the foot-base and give an indication of the percentage contribution of each of these factors.

The use of a force plate allowed direct measurement of the horizontal moment at

the feet. The means were somewhat larger than those calculated in Chapter 3 for exertions under similar conditions (70Nm vs 41Nm for exertions to the right, and 69Nm vs 38Nm for exertions to the left). However, the forces exerted by the subjects in this experiment (see section 4.5.3, p.150) were slightly less than the means of the forces (as a percentage body weight) exerted by the same group of subjects as in Chapter 3 during strength measurement trials by Pinder *et al.* (1995), rather than those measured while participating in the photographic session of Chapter 3. As Figure 3.9 (p.89) showed, the forces exerted during strength measurement were usually greater than those measured in the photo session.

The magnitudes of the horizontal moment generated at the feet were almost identical for exertions to the right and the left (approximately 70Nm), even though the forces exerted as a percentage of body weight differed by 10%. This suggests that the position of the feet was changed so that the horizontal distance between the centre of foot pressure and the hand increased to provide a greater counteracting moment at the foot-base for exertions to the left.

Because only one force plate was available, it was necessary to make the assumption that the conditions of all three situations during the experiment (i.e., the left, the right, or both feet on the force plate) were similar. The high correlation coefficients obtained between the sums of the variables obtained when the left or right feet were on the force plate and those obtained when both feet were in contact strongly support this assumption.

4.8 SUMMARY OF SURFACE ELECTROMYOGRAPHY AND FORCE PLATE EXPERIMENTS.

4.8.1 Summary of results.

The results from both the surface electromyography and force plate experiments concerning exertions to the right and to the left are summarized in Table 4.9 and Figures 4.17 and 4.18. The emg scores for the left lower limb are derived from those measured in the right lower limb when exerting in the opposite direction with the left hand.

Table 4.9 Summary of results. Activity of lower limb muscles and the percentage contribution to the overall horizontal moment by different factors during exertions to the right and left. D - direction of exertion, (ll) - lower limb, L - to the left, R - to the right, (l) - left lower limb, (r) - right lower limb, Q - quadriceps femoris, H - hamstrings, T - tibialis anterior, G - gastrocnemius, Mch A - Mechanism A, Tz(L) - pure torque exerted by left lower limb, Tz(R) - pure torque exerted by right lower limb.

	Surface emg				Force plate		
D (ll)	Q	H	T	G	Mch A	Tz(L)	Tz(R)
R (l)	1	4	1	2	48	37	15
R (r)	4	2	4	3			
L (l)	3	1	2	1	52	16	32
L (r)	2	4	2	2			

4.8.2 Discussion.

The surface emg and force plate experiments were an attempt to elucidate some of the mechanisms by which the lower limbs transmit both a horizontal thrust and torque to the floor at the same time. It was shown in Chapter 3 that the lateral force at the hand, acting at some distance from the centre of foot pressure, results in a horizontal moment. Since the exertions are all static, this must be counteracted by a horizontal moment at the foot-base in the opposite direction. The force plate experiment has shown that approximately half of this moment is generated through the action of one lower limb working parallel, but in the opposite direction, to the other. Of the remaining half, approximately two thirds are due to pure torque exerted by the right lower limb when exerting to the left and by the left lower limb when exerting to the right. The last third is pure torque exerted by the right lower limb when exerting to the right and the left lower limb when exerting to the left.

The surface emg experiment has given some insight into the muscular demands on the lower limbs during lateral exertion. Although the experiment studied only the flexors and extensors of the lower limb joints, the idea that the flexors of the hip and extensors of the knee of one lower limb were active at the same time as extensors of the hip and flexors of the knee of the other was supported.

Although not done in this study, investigation of the abductors and adductors might give an indication of how these muscles help to provide a counteracting lateral

force at the foot-base. Surface emg of the deep lateral rotators of the hip is not possible due to their inaccessible position in the gluteal area, but if it were possible to discover their activity during exertion to the left or right, it might shed light on the way the muscles of this region are used in these manoeuvres.

4.9 CONCLUSIONS.

1. The suggestion that the horizontal moment generated at the feet during exertions to the left or right is partly produced by one lower limb working in parallel with, but in the opposite direction to, the other has been supported by the results of both the surface electromyography and the force plate experiment.
2. The suggestion that the horizontal moment is partly produced by individual torques of each lower limb has also been supported by the results of the force plate experiment.
3. The magnitude of the contribution to the horizontal moment is approximately one half by the action of one lower limb exerting a force in a direction opposite to the other, one third by the right lower limb when exerting to the left or the left limb while exerting to the right, and one sixth by the left lower limb when exerting to the left or the right limb when exerting to the right.
4. The use of the force plate allowed direct measurement of the horizontal torque, which reached a mean of 70Nm for exertions to the right and left.

CHAPTER 5.

GENERAL DISCUSSION.

5. GENERAL DISCUSSION.

This work has presented an investigation into some novel aspects of static asymmetrical whole body exertion while standing. Although a few studies of asymmetrical exertion have been carried out, none have described the finding of a horizontal turning moment at the foot-base, which is of a magnitude that would have important implications for the exertion of forces with a lateral directional component.

In Chapter 3 of the thesis, the horizontal moment was deduced from the results of a detailed postural analysis of subjects exerting in many directions both in and out of the fore-aft plane. Although the magnitudes of forces in directions with a lateral component were often modest, the magnitudes of corresponding moments (mean of eleven subjects) sometimes reached as high as 30Nm.

The large data set showing postures adopted during one-handed exertions in several directions demonstrates that similar strategies for exerting force were used by many subjects. A detailed comparison of the adopted postures with strength showed that greater forces were achieved by both better deployment of body weight (*dead* component) and increased muscular effort (*live* component). It is likely that motivational factors play a part in both of these strategies. Previously, the *dead* component of an applied force has frequently been described as being due to the deployment of body weight (Grieve and Pheasant, 1982; Grieve, 1983a). However, when a person is exerting downward from above or below a handle, the centre of

foot pressure may be (horizontally) close to the hand centroid with the centre of gravity almost vertically above it. In these cases, the slope of the equation of static exertion (see section 3.1, p.71) would be close to vertical and the *dead* component of force small (as seen in some subjects in this study), but the exertion could still be described as efficient deployment of body weight. It is therefore proposed that the *dead* force is better described as being due to the *horizontal* deployment of body weight in relation to the centre of foot pressure.

The experiments of Chapter 3 have shown that the Postural Stability Diagram is a valid method of static task analysis regardless of the direction of exertion. The information gained about relationships between posture and strength may be considered for use in worker training. It appears that in many people, it is not instinctive to make the best use of body weight in order to exert greater forces. This was also demonstrated by the subjects of Chaffin *et al.* (1983) for pushing and pulling, who did not move their feet backward as far as possible for pushing or forward for pulling, despite the fact that this would have led to greater forces being exerted. Admittedly, their subjects were untrained college students rather than industrial workers, but the same was found in the present study where the subjects were army personnel, more used to heavy manual tasks.

In these experiments, the foot position was restricted so that each foot was equidistant from the central longitudinal line beneath the handle. The subjects exerting greater lateral forces had the centre of foot pressure closer to one foot. As with pushing and pulling, the further apart the centre of gravity and the centre of

foot pressure, the greater the force exerted (from the Postural Stability Diagram).

In an industrial situation, where space restrictions may prevent a person from pushing a handle from behind, greater forces could be exerted if, for a force to the left, he or she placed the right foot as far to the right as possible. However, to prevent stability being compromised, a wide stance with the left lower limb providing support in case of a slip would be preferable.

The surface electromyography readings from quadriceps femoris and the hamstrings were consistent with the muscle activity expected if one foot exerted a static forward force on the floor at the same time that the other exerted a backward force (Mechanism A). Tibialis anterior showed more activity when the lower limb was exerting a forward force due to Mechanism A, but was also active when subjects pushed horizontally forward. The foot is exerting a forward force upon the floor in the former case, and a backward force in the latter. This suggests that the anatomical strategies differed depending on the direction of exertion. The gastrocnemius showed no convincing pattern for lateral exertions, which suggests it is not implicated in either mechanism for producing horizontal torque.

Biomechanical models that predict strength (Chaffin, 1969; Chaffin and Baker, 1970; Martin and Chaffin, 1972; Garg and Chaffin, 1975) presume that whole body strength is limited either by compressive stress on the lumbar spine, or by the maximal voluntary torque of a particular joint, which has been previously measured in a specific direction with the joint in a fixed posture. The models do not take into account the requirement of some muscles to be employed in the generation of

a horizontal turning moment at the foot-base, which may reduce their ability to exert forces for other purposes. Maximal torques for medial and lateral rotation of the lower limb are not included in the models.

The force plate allowed the magnitude of the horizontal torque to be measured directly. Torque produced during (dynamic) axial rotation of the trunk has been measured previously (Grieve and Arnott, 1970), and the mean (with feet apart) was found to be 70Nm, which is identical to the mean moment measured in Chapter 4. In this study, it was found that approximately half of this was due to one foot exerting on the floor in a direction opposite to that of the other (Mechanism A). In the experiments of Grieve and Arnott, the mean torque was found to increase from 48Nm when the subjects stood with their feet together, to 70Nm when the feet were apart. Although these figures were measured under dynamic conditions, the subjects exerting greater torques were likely to be utilizing Mechanism A. When the feet were close together, the distance between the forces acting in opposite directions was less, and therefore the torques generated were smaller. The mean torque exerted with feet together was 69% of that with the feet apart, which suggests that at least some part of the horizontal moment was generated by Mechanism A.

The second method of generation of horizontal moment was due to individual torque applied by each foot separately upon the floor, with the greater percentage, not surprisingly, generated by the foot bearing more of the body weight. (The centre of foot pressure was usually closer to one foot, since a lateral force was

simultaneously being applied to the floor to counteract the force at the hands - see Figures 3.10, p.91 and 3.15a, p.97).

Limitations in a person's ability to exert forces include poor coupling at the interfaces between the hand and the handle and between the feet and the floor (as discussed in Chapter 2). In all the experiments conducted in this study, these factors were considered. The handles conformed to the specifications suggested by Drury (1985), with diameters of 35mm, lengths between 112mm and 136mm and clearances of over 30mm all round.

The platforms on which the subjects stood were covered in emery cloth, and the subjects wore trainers or similar rubber-soled shoes, so that the limiting coefficient of friction was maximized. No slipping occurred. In an industrial environment, it is unlikely that such high coefficients will be present. Horizontal torques at the floor in relation to slipping has been considered by Grieve (1983b), who states, "If the person attempts to exert torque at the feet in the horizontal plane, the contact area of the foot base will reach a slip condition due to translational forces sooner than would be reached without the torque". This is because any point on the sole of a shoe is able to withstand a force that is dependent on the coefficient of friction and the pressure at that point. The force may be due to translation, torque or (as in this study) both. "It follows that an attempt to transmit torque in the plane of the floor will place demands upon the shoes which detract from the capacity to transmit translational forces (and *vice versa*)". Therefore a person exerting lateral forces would require a higher limiting coefficient of friction of the floor than if

exertions were carried out in the fore-aft plane. To resist slip, Grieve suggests a wide stance (which would facilitate the generation of torque by Mechanism A) with the force equally distributed between the feet. The latter condition would be more difficult to achieve, as this study has shown that during laterally directed exertions, the centre of foot pressure is closer to the foot that also generates a greater individual torque. However, the normal force at this foot (and therefore the tangential force required for slip to occur) would be greater than that at the other foot, which generates a smaller horizontal torque.

5.1 FUTURE WORK.

This study has examined the role of the flexors and extensors of the lower limb in the generation of horizontal torque at the foot-base. However, it is likely that the co-ordination of the muscles of the lower limb during laterally directed exertions is complex, since, as previously mentioned, their role is threefold - to support the weight of the body, to exert a lateral force to counteract that at the hands, and to generate the horizontal moment. It would be useful to examine the activity of other muscle groups during lateral exertions. A future experiment might combine several of the techniques used in this thesis. Electromyography of several muscles of the lower limb could be combined with recording of posture, measurement of the force at the hands and use of two force plates. This would allow a biomechanical analysis to be performed that would provide details of the forces and moments at each joint, including those for the joints of each lower limb separately. Obvious muscles to be investigated are the abductors and adductors of the hip, because of

their likely involvement in the production of a lateral force upon the floor. The maximal torque of medial and lateral rotation of the leg, with contribution by the muscles of the thigh restricted, has been measured previously at 31Nm and 20Nm respectively with the hip flexed 10° and the knee flexed 20° (Shoemaker and Markolf, 1982). This suggests that the muscles of the leg are at least capable of generating a horizontal torque. In this study, only tibialis anterior and gastrocnemius were investigated, but other muscles may play a role in its generation.

Although fine wire electromyography may be used to investigate deeper muscles, this would be dangerous in the case of the deep lateral rotators of the hip and no other sure method of judging the relative activities of deep muscles has been found. Magnetic resonance imaging has been used to measure the moment arms of muscles with the ankle joint in different positions (Rugg *et al.* 1990) and it may prove useful in the measurement of muscle lengths during specific movements of joints. For example, the length of some of the deep lateral rotators may be measured while in the anatomical position and again when the lower limb has been laterally rotated. This would give greater insight as to their action *in vivo*, but would unfortunately be of limited value in investigating muscle activity under isometric conditions, as in this study.

Other future experiments might be conducted with subjects in fixed postures. Standing upright would simplify the discrimination of muscle activity due to postural demands and due to the exerted force. The effect on Mechanism A of

placing the feet together or fixed distances apart in the frontal and sagittal plane during lateral exertions could also be examined.

Most exertion in a work environment is performed under dynamic conditions, whereas this study has been concerned only with the static case. The study of dynamic lateral exertions requires the consideration of inertial and accelerative properties of the load. Although Grieve and Arnott (1970) have investigated torque production during axial rotation of the trunk, horizontal moments at the feet were not specifically targeted. As with static three-dimensional biomechanical models, dynamic models could also incorporate the moment as part of the analysis.

5.2 GENERAL CONCLUSIONS.

1. For one-handed exertions in directions both in the fore-aft plane and with a lateral component, greater forces can be achieved by better horizontal deployment of body weight as well as increased muscular effort, particularly in directions with a large horizontal component.
2. The Postural Stability Diagram and the equation of static exertion may be used in the analysis of exertions in any plane.
3. Exertion in directions with a lateral component is accompanied by the generation of a horizontal turning moment at the foot-base, which may be substantial when the applied forces are large.
4. Approximately half of the horizontal moment is produced by one foot exerting a force on the floor in the opposite direction to the other foot, and one half is produced by the feet exerting torque upon the floor individually. Of this latter half, approximately two thirds are generated by the left lower limb when exerting to the right or the right lower limb when exerting to the left, and the remaining third by the right lower limb when exerting to the right or the left lower limb when exerting to the left.
5. The quadriceps and tibialis anterior are active when the lower limb exerts a forward force on the floor in the generation of the horizontal moment during lateral exertions, while the hamstrings are active when a backward force is exerted upon the floor.

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CHAPTER 6.

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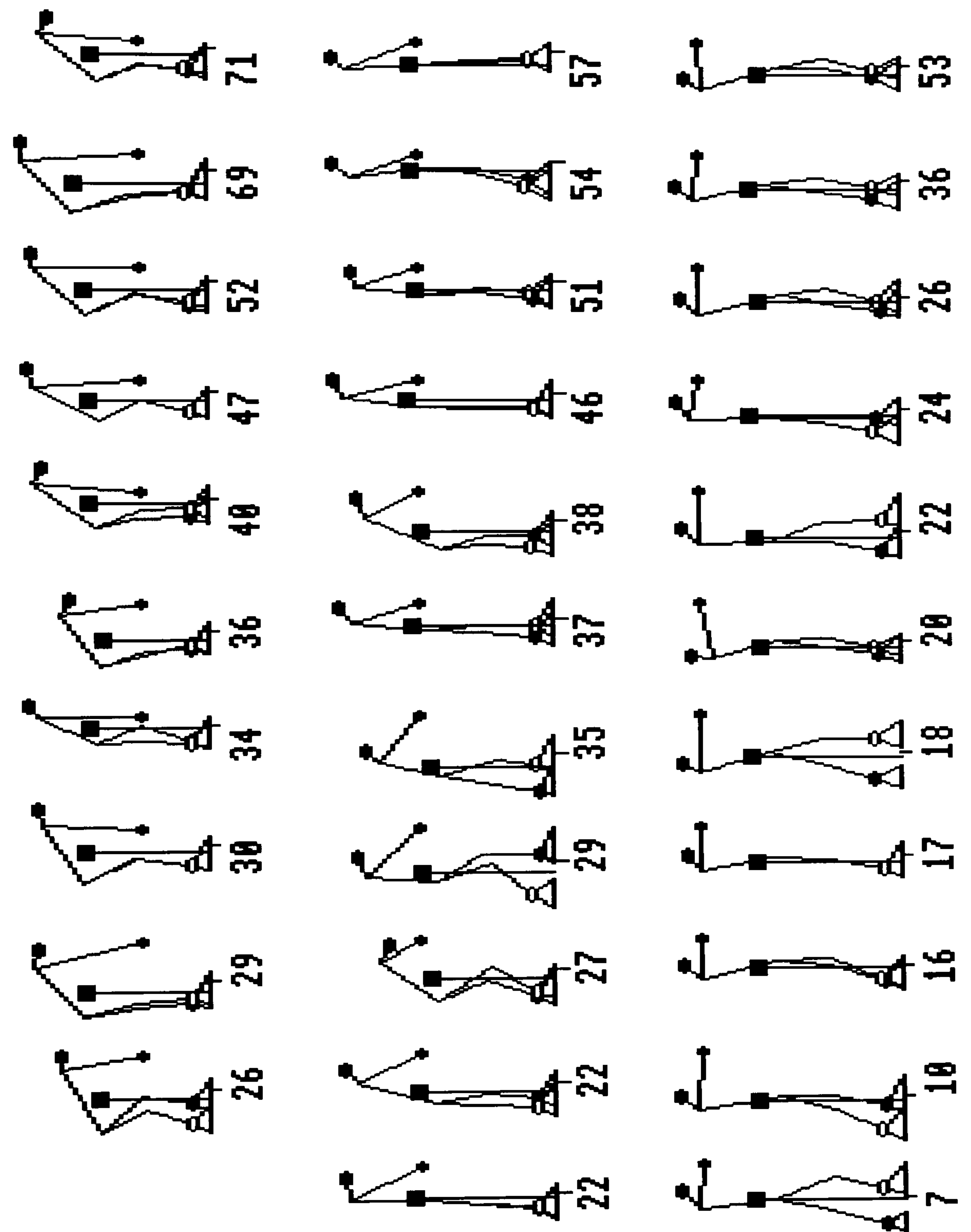
APPENDICES.

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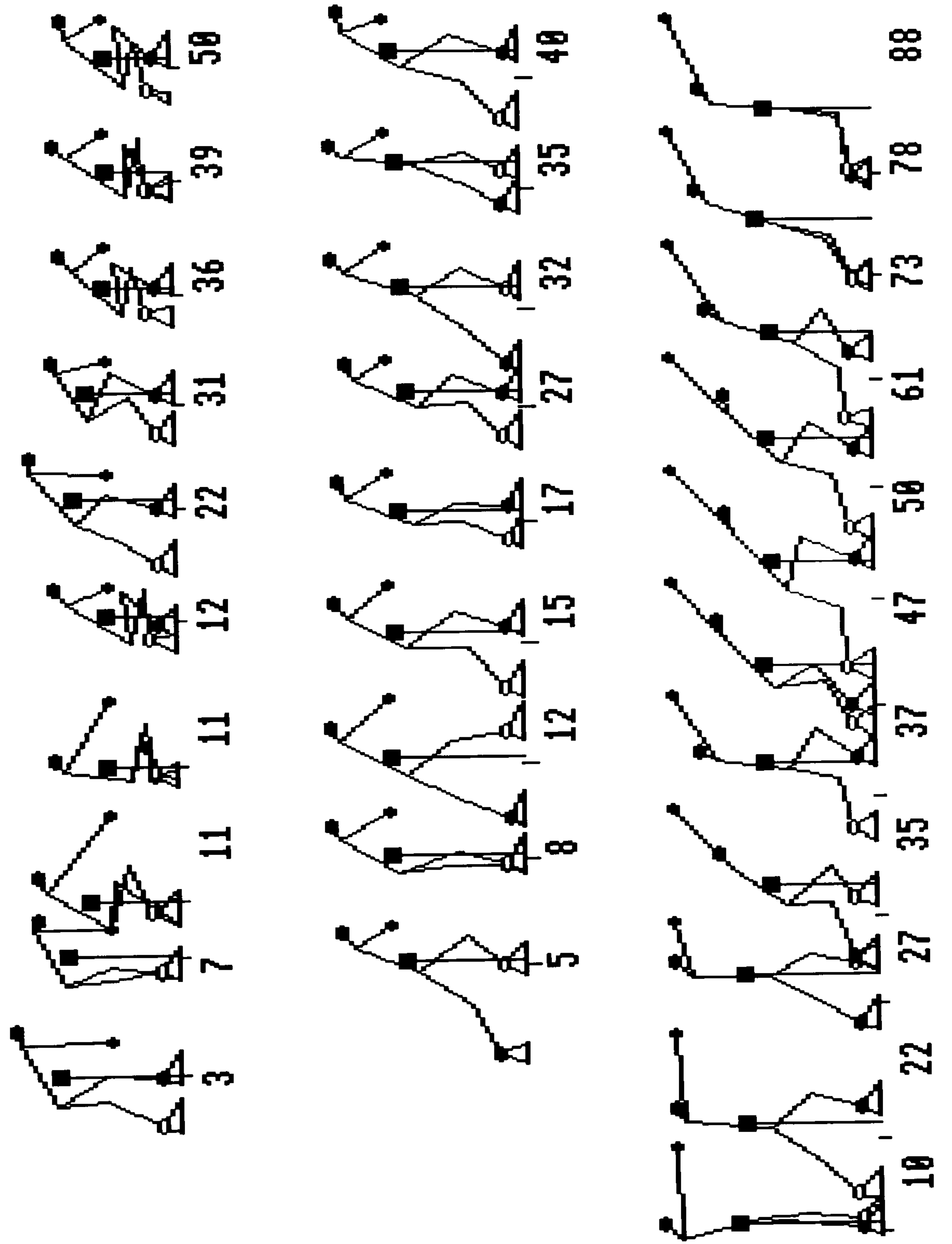
APPENDIX A.

Complete set of postures of subjects from Chapter 3 at the three handle heights, showing force exerted as a percentage body weight.

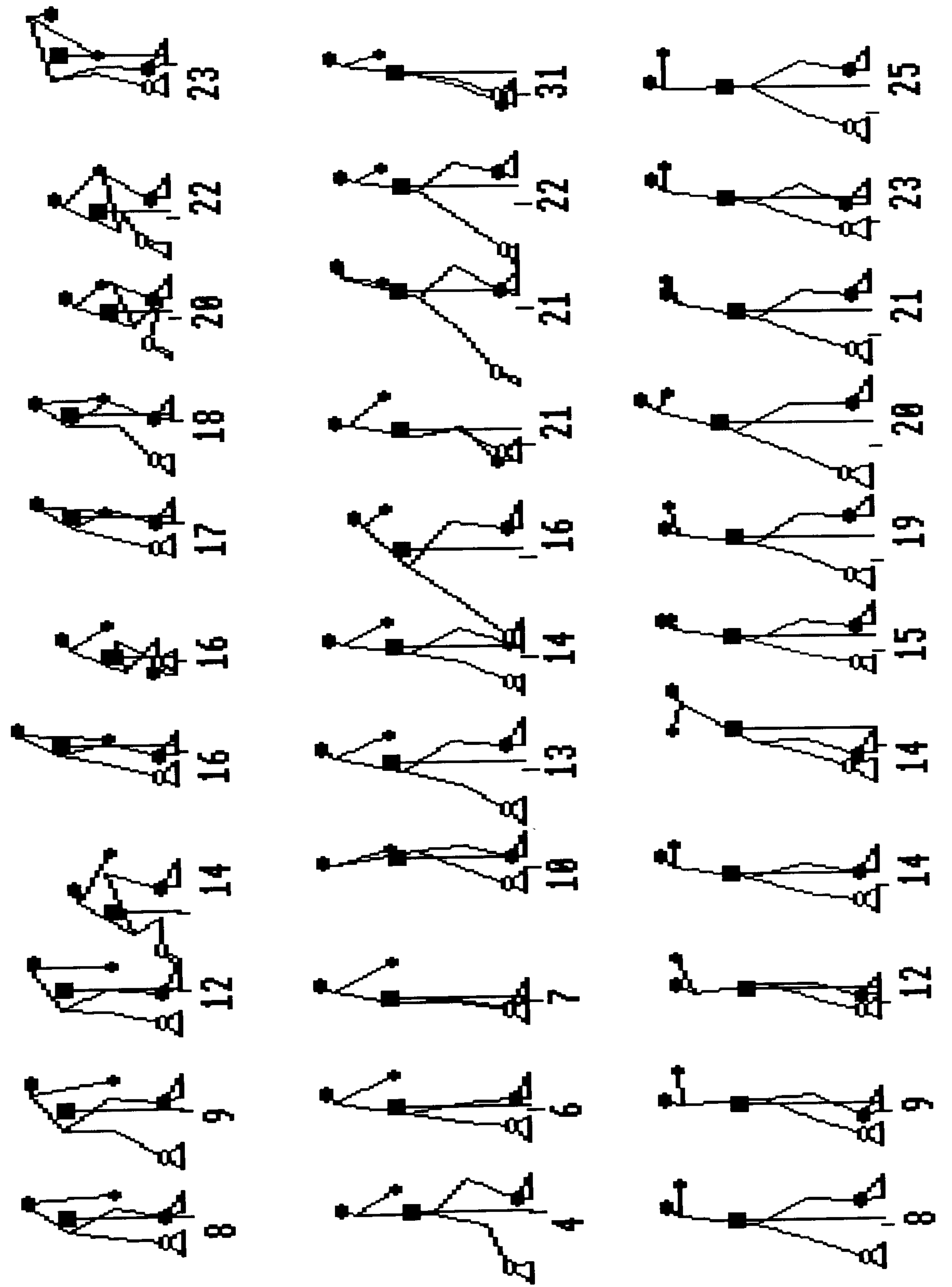
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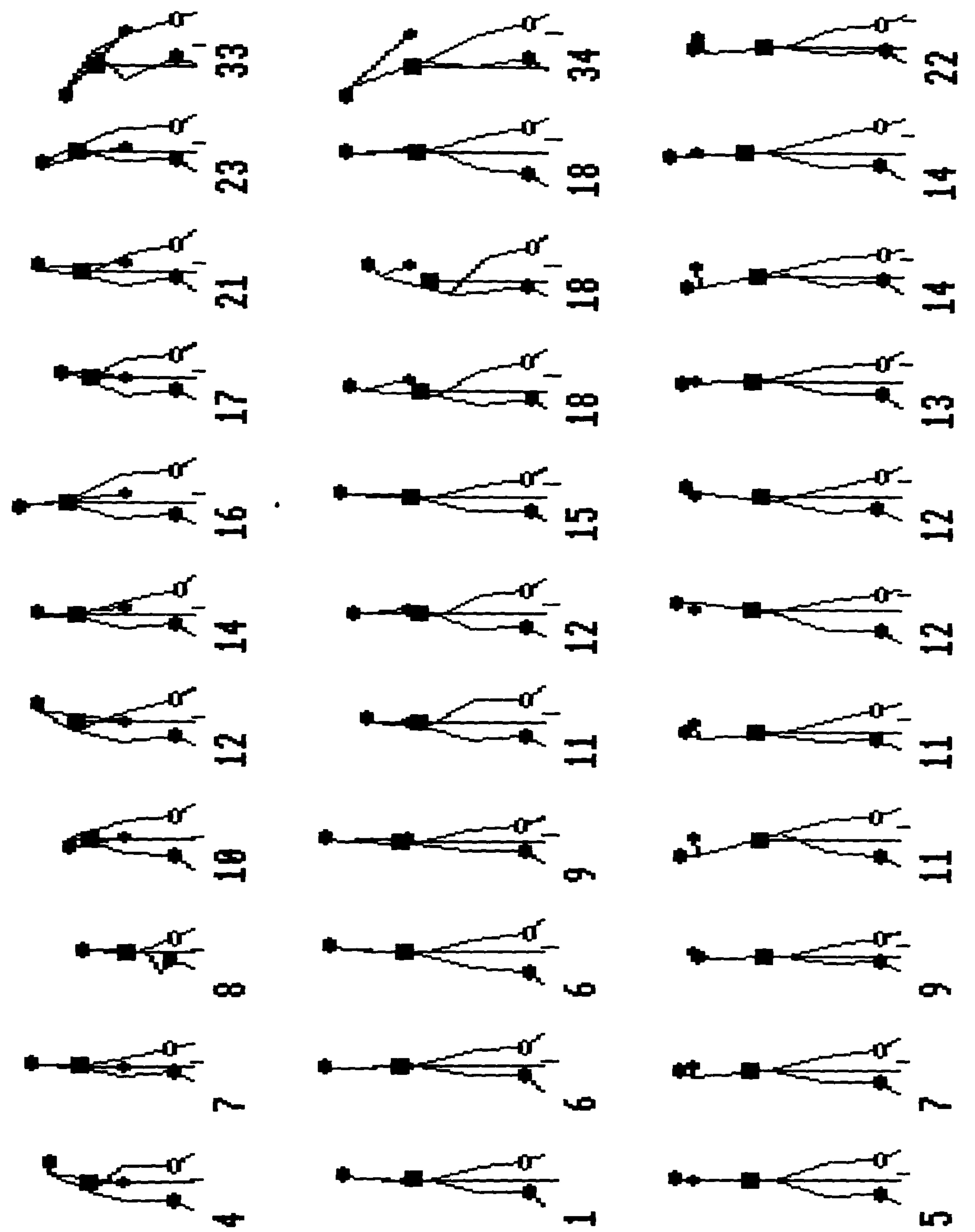
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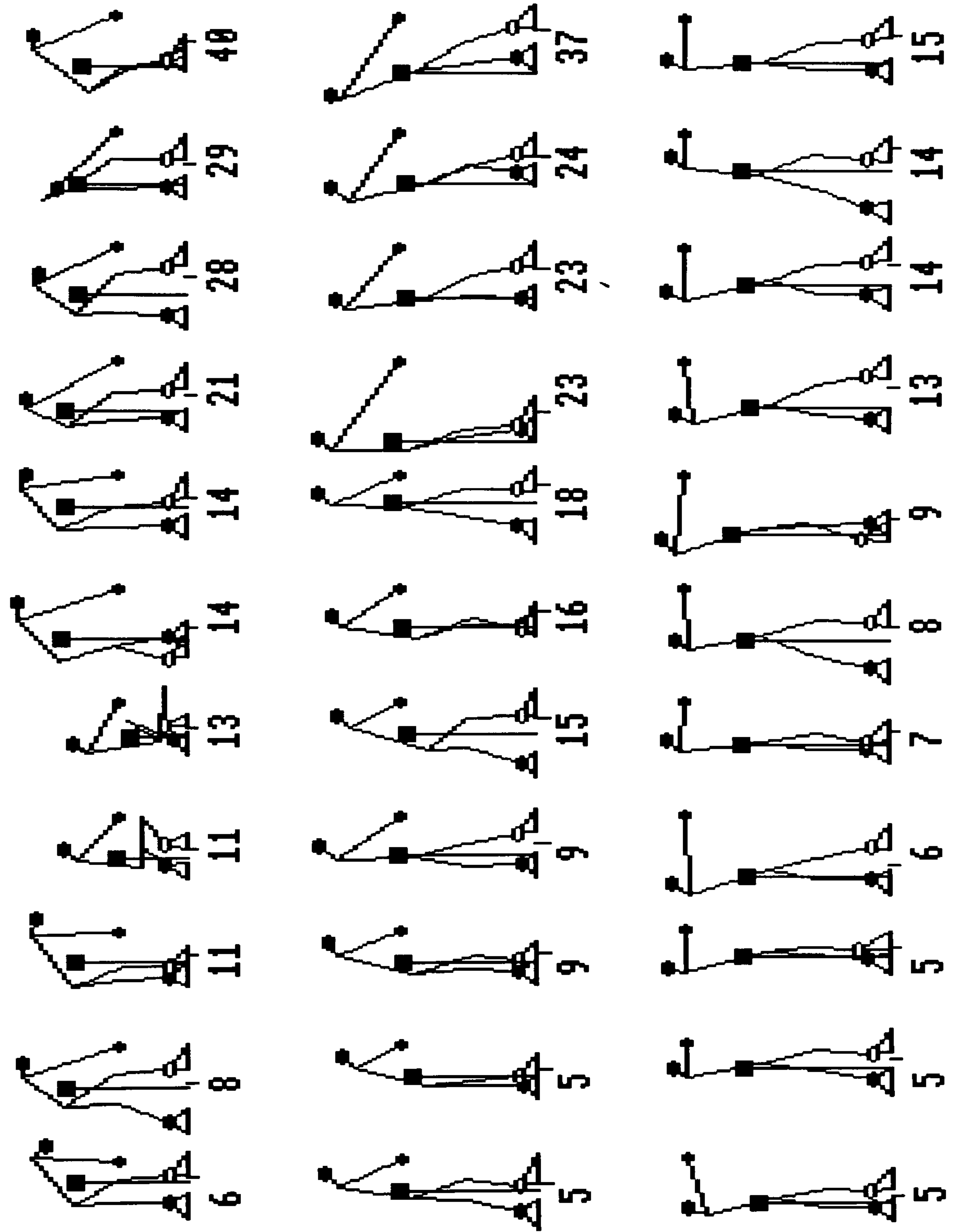
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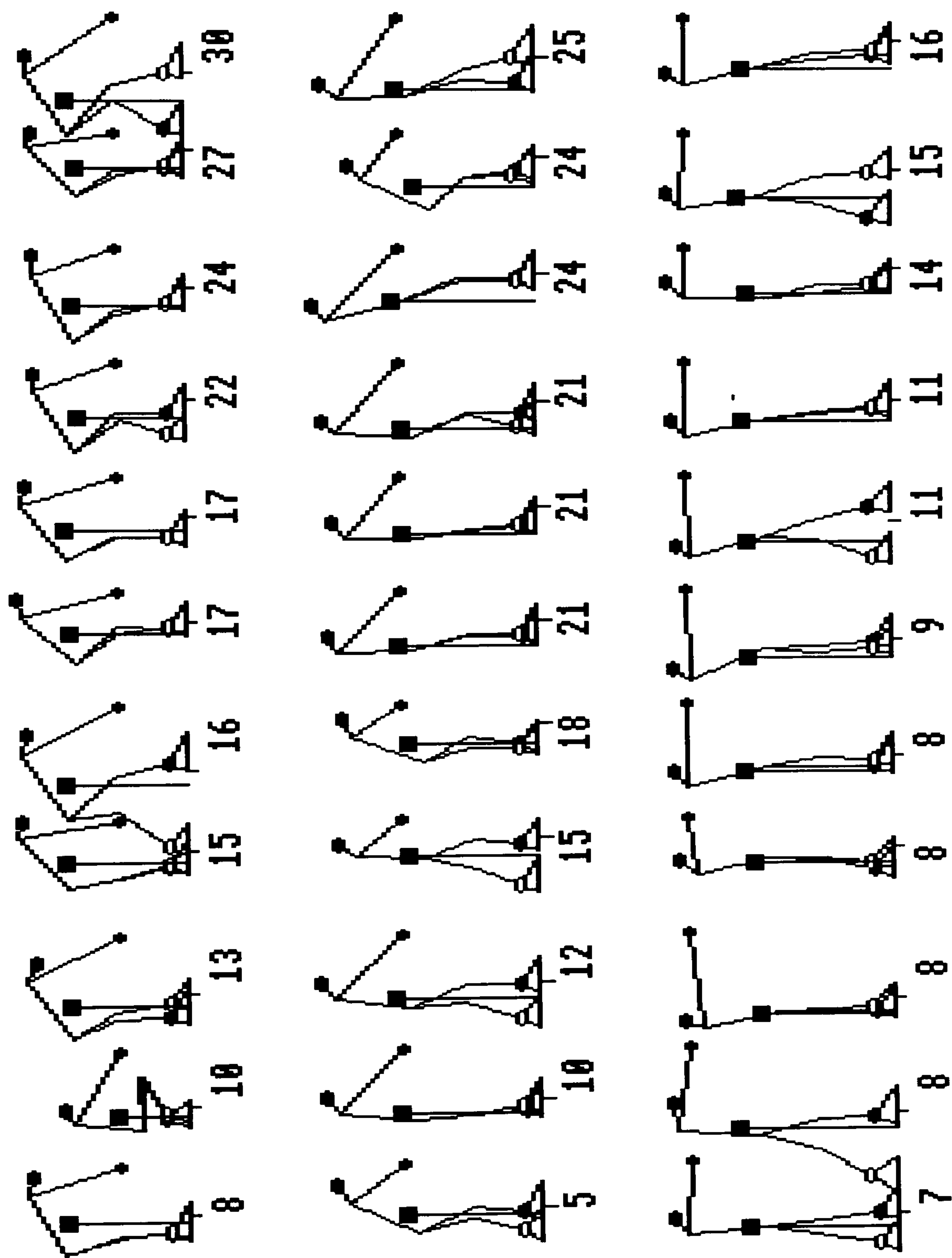
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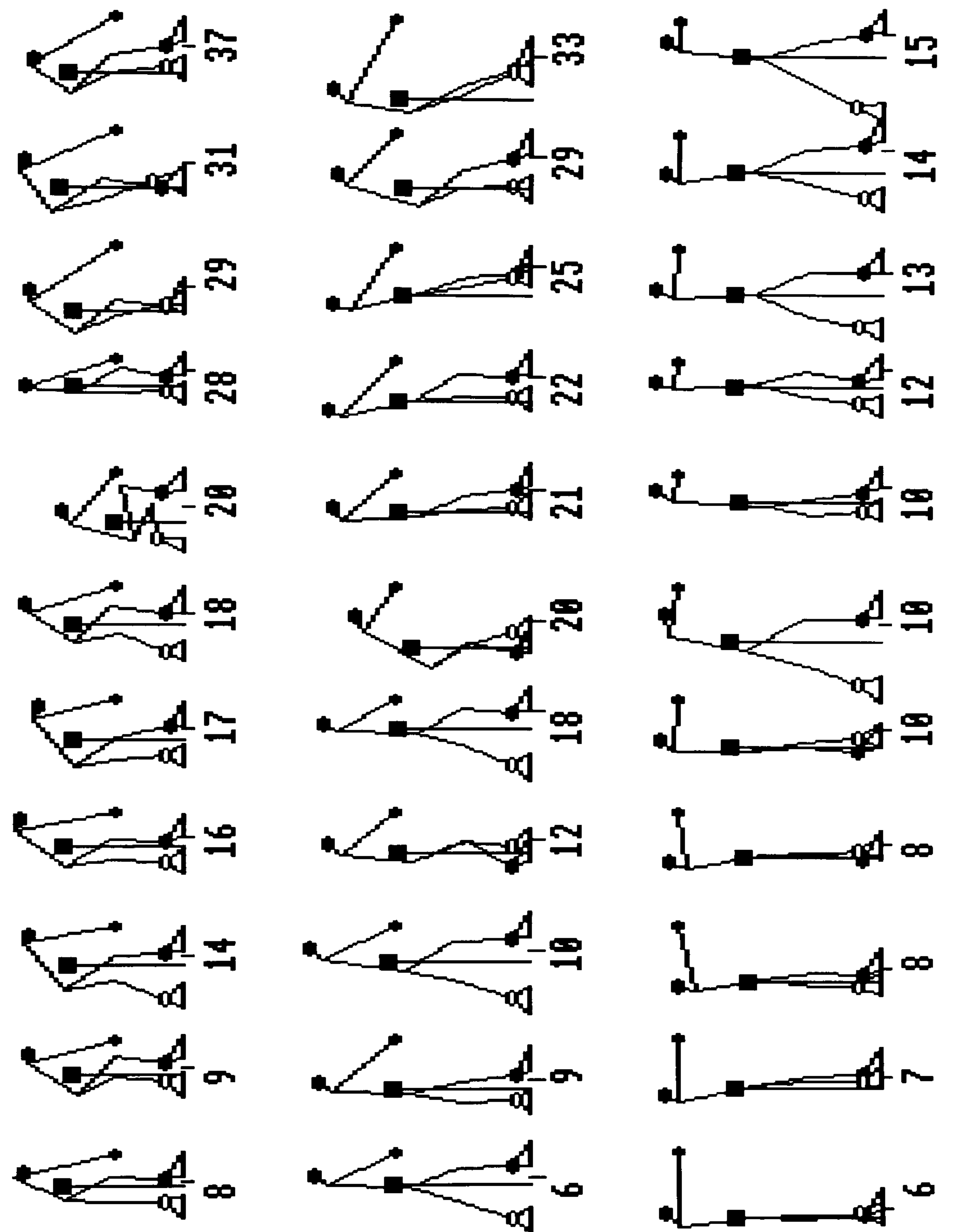
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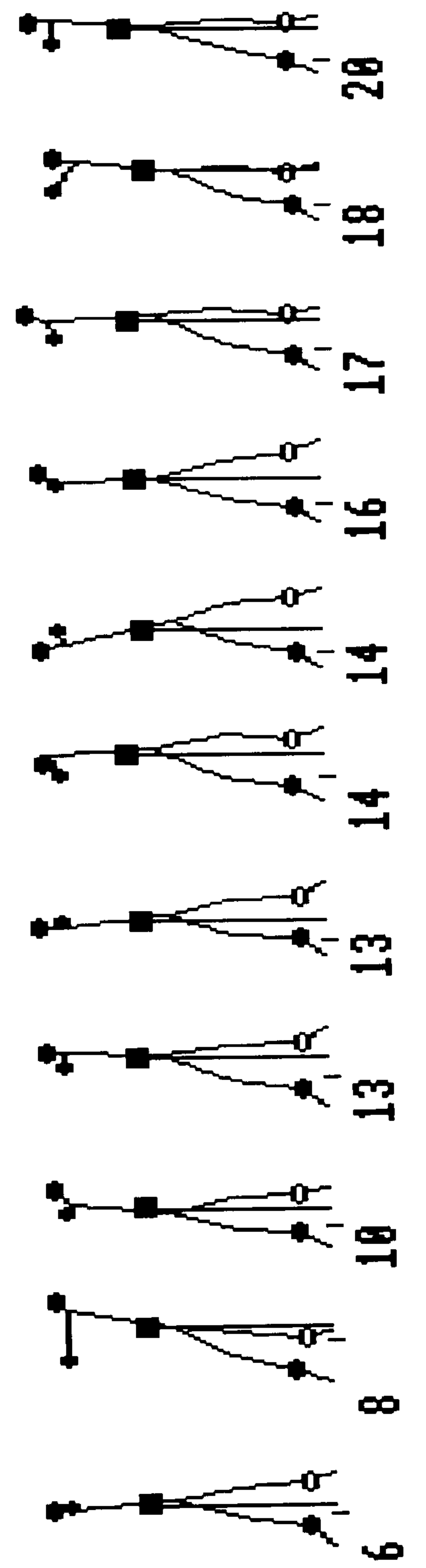
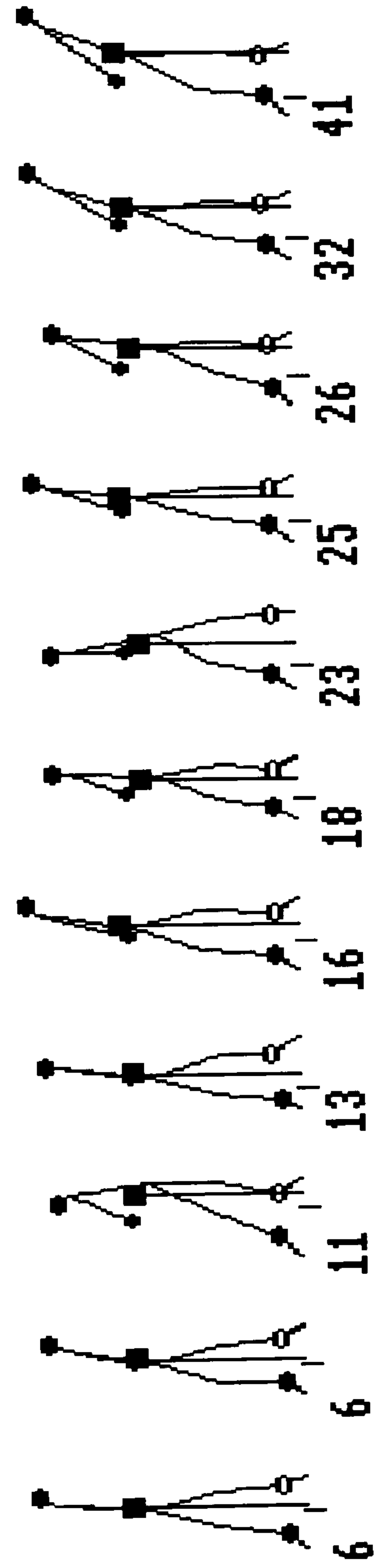
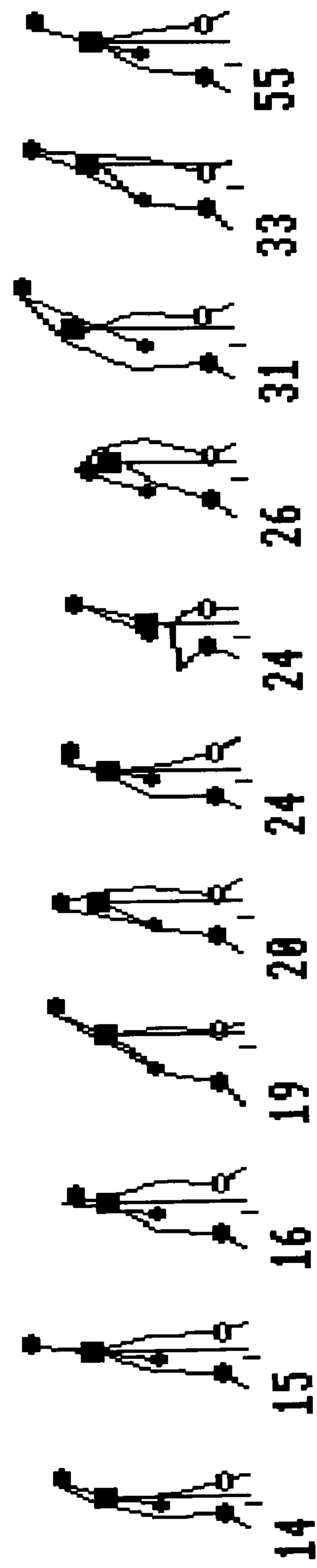
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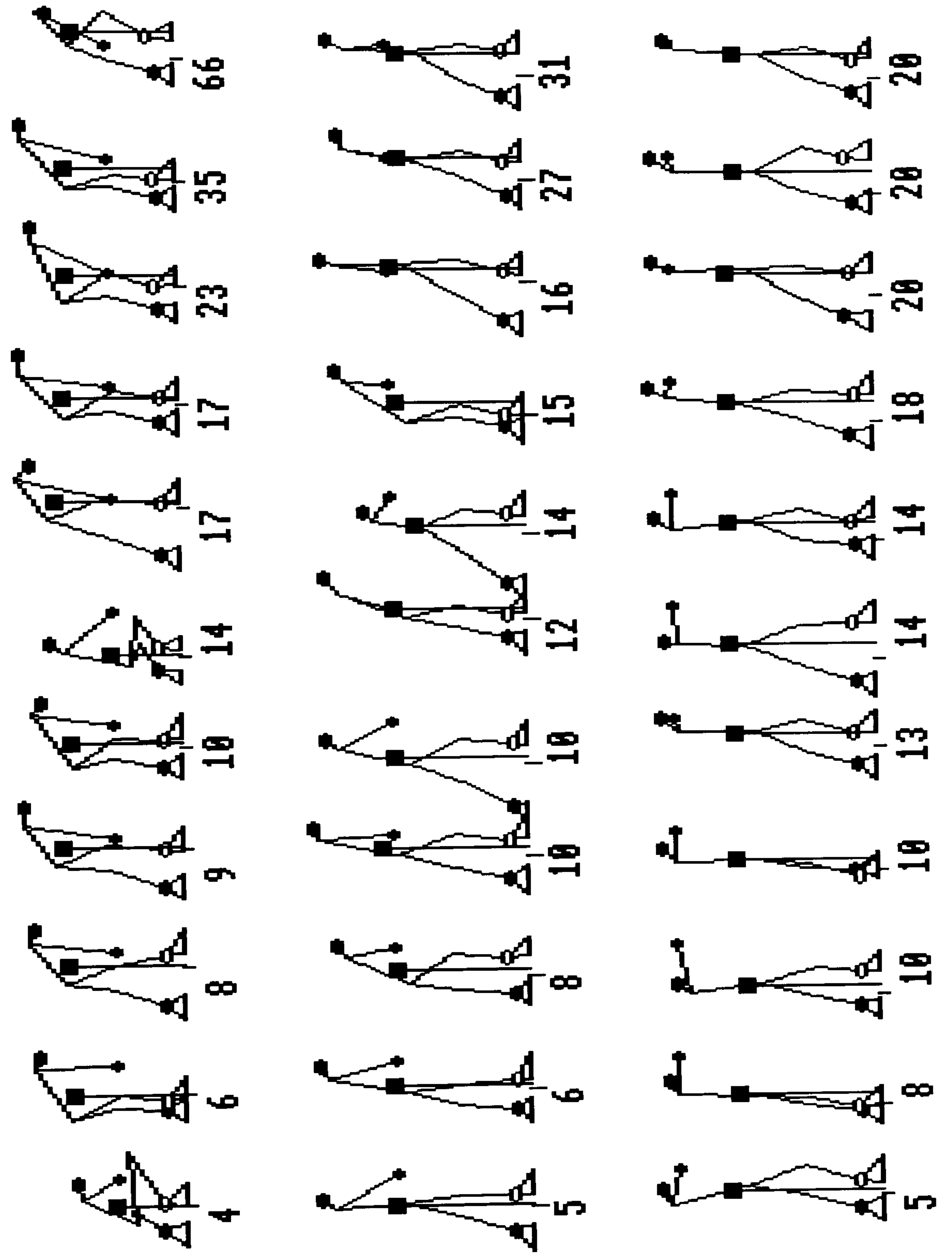
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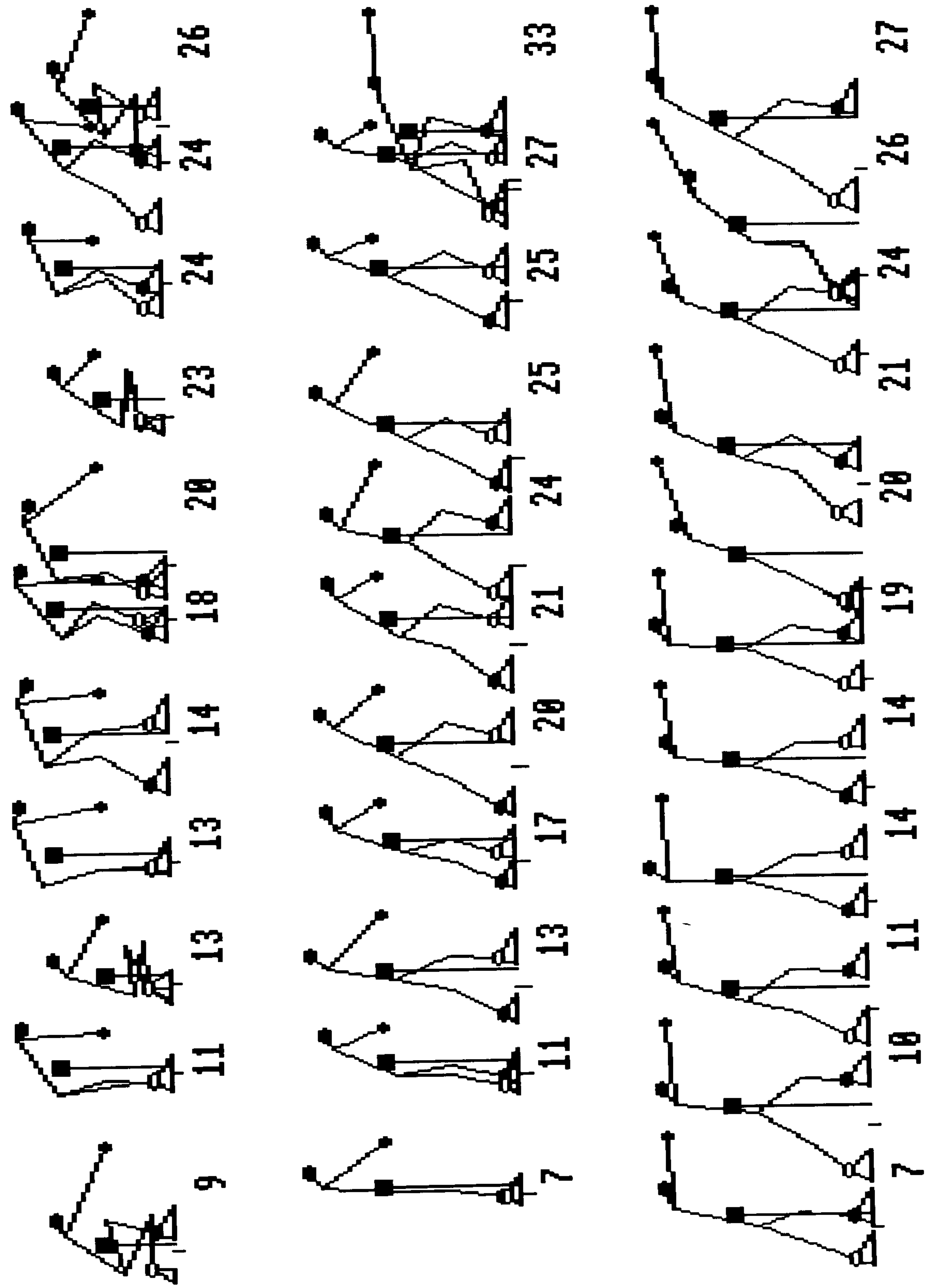
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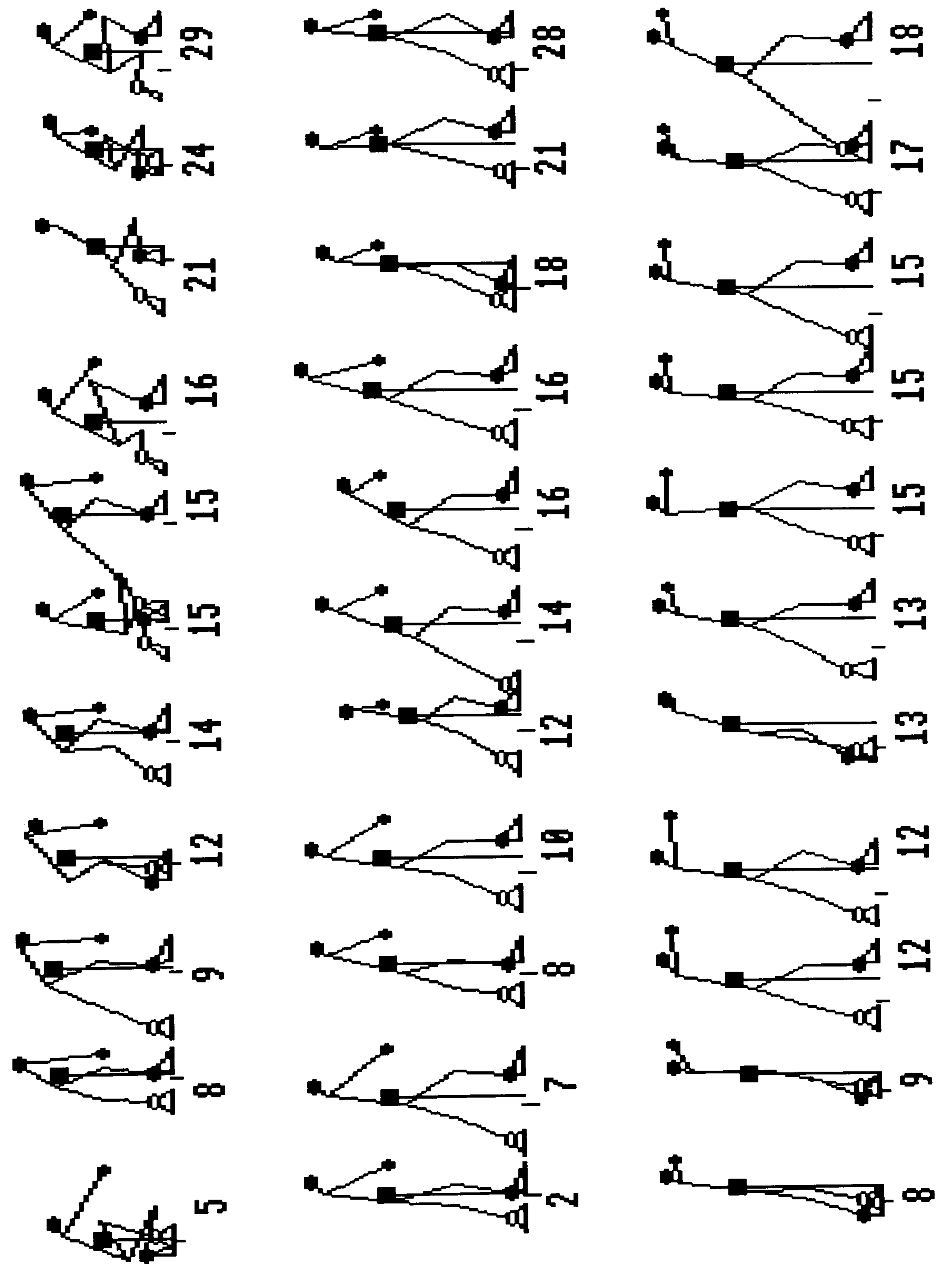
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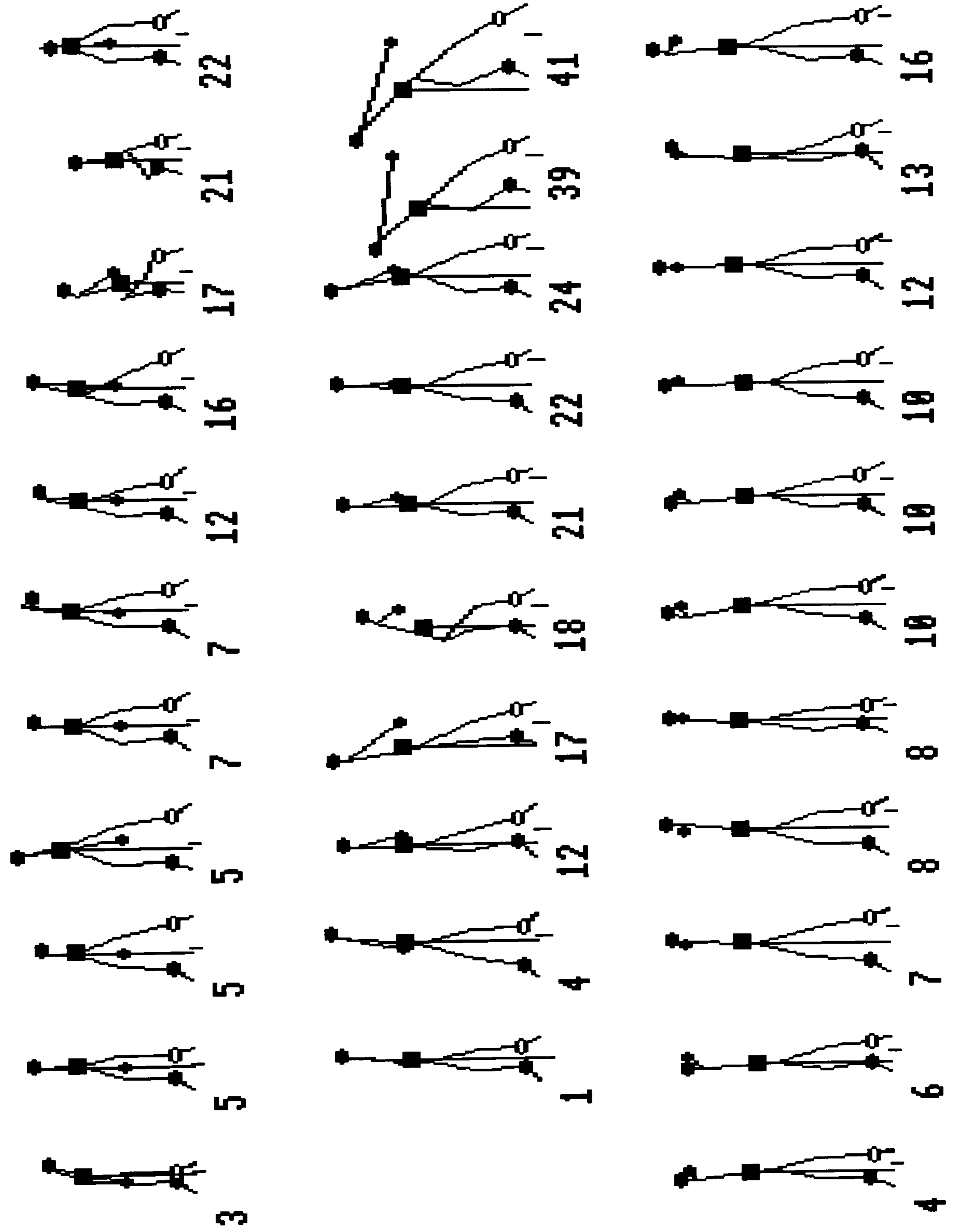
DIRECTION OF EXERTION: HORIZONTAL AND FORWARD



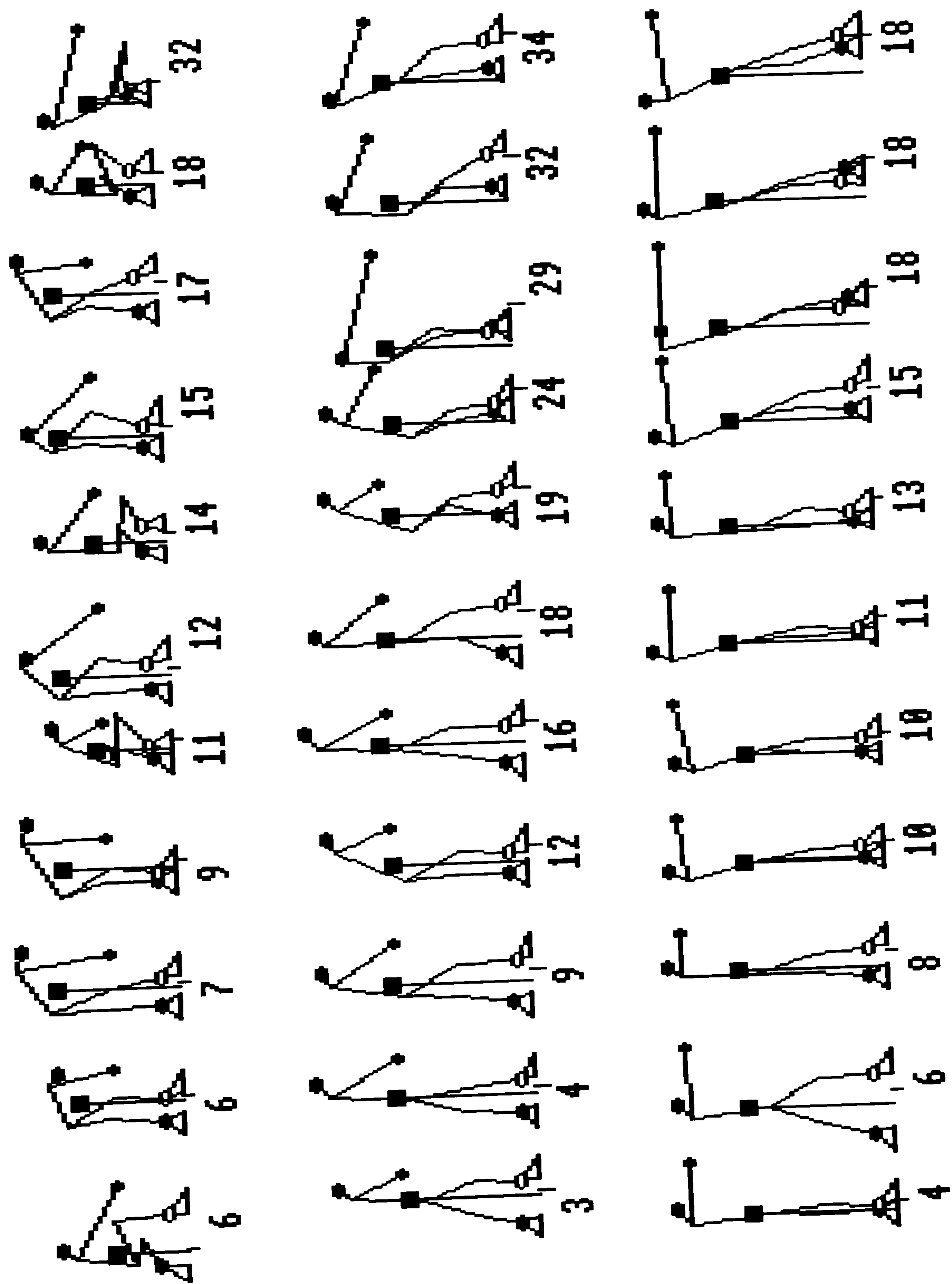
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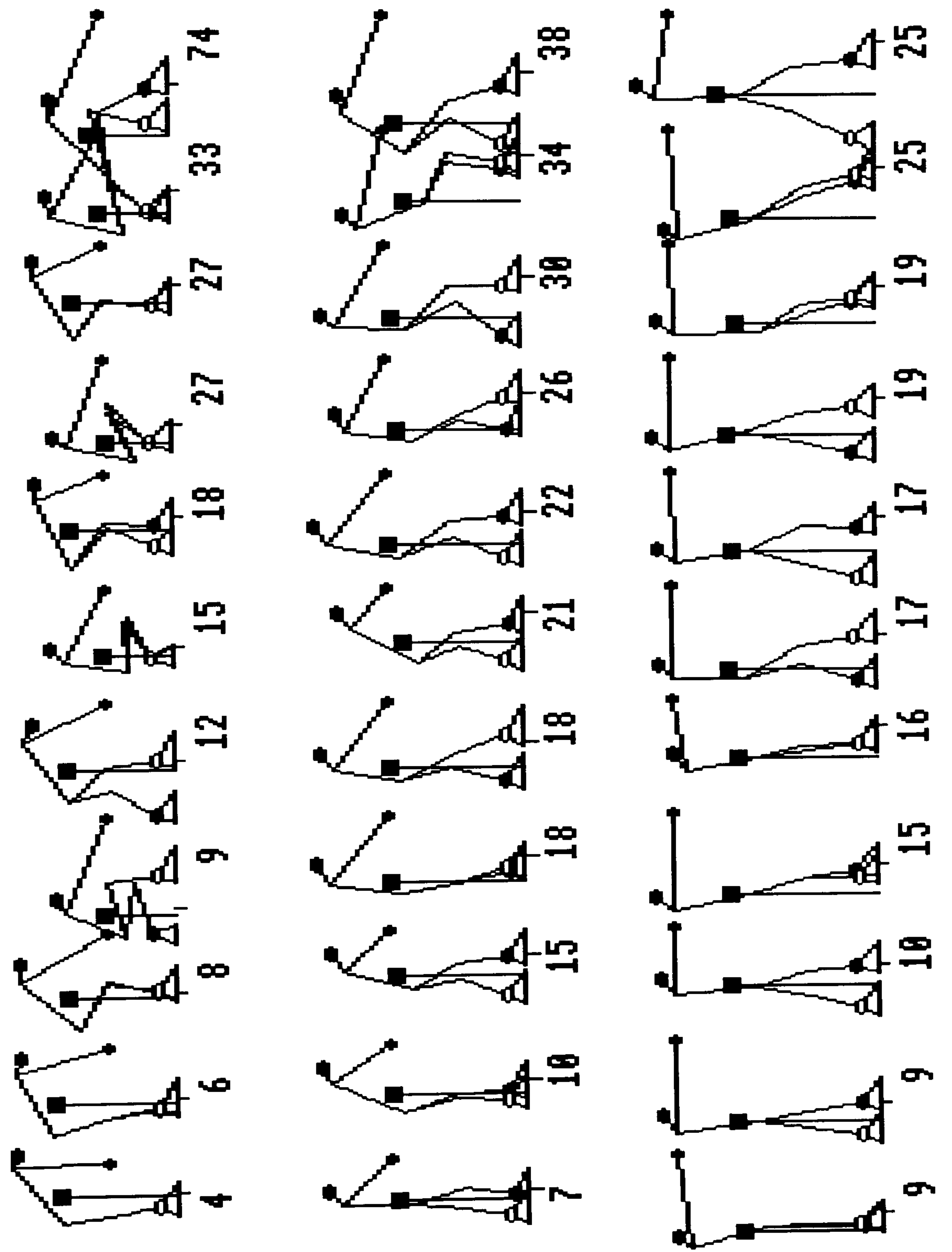
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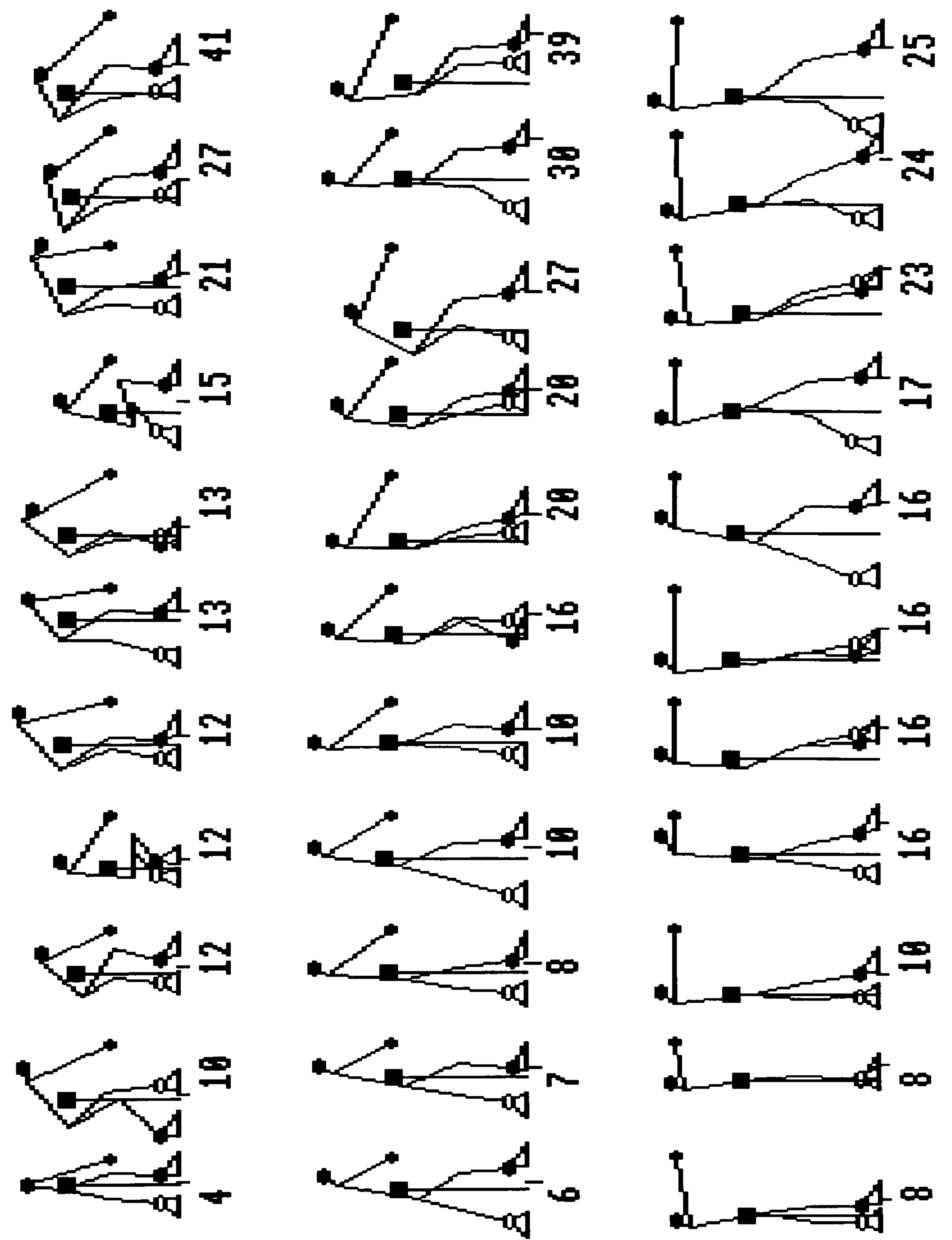
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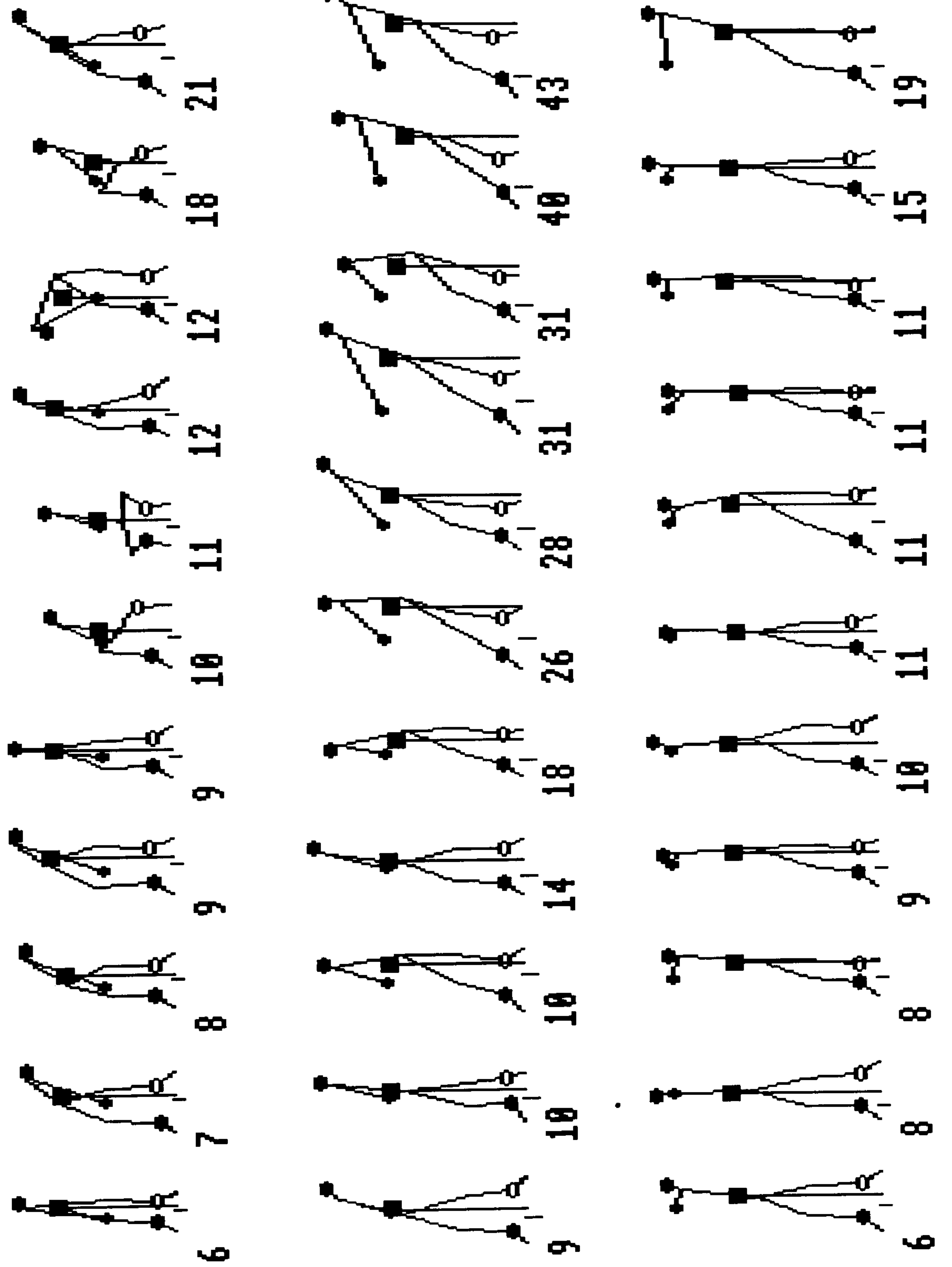
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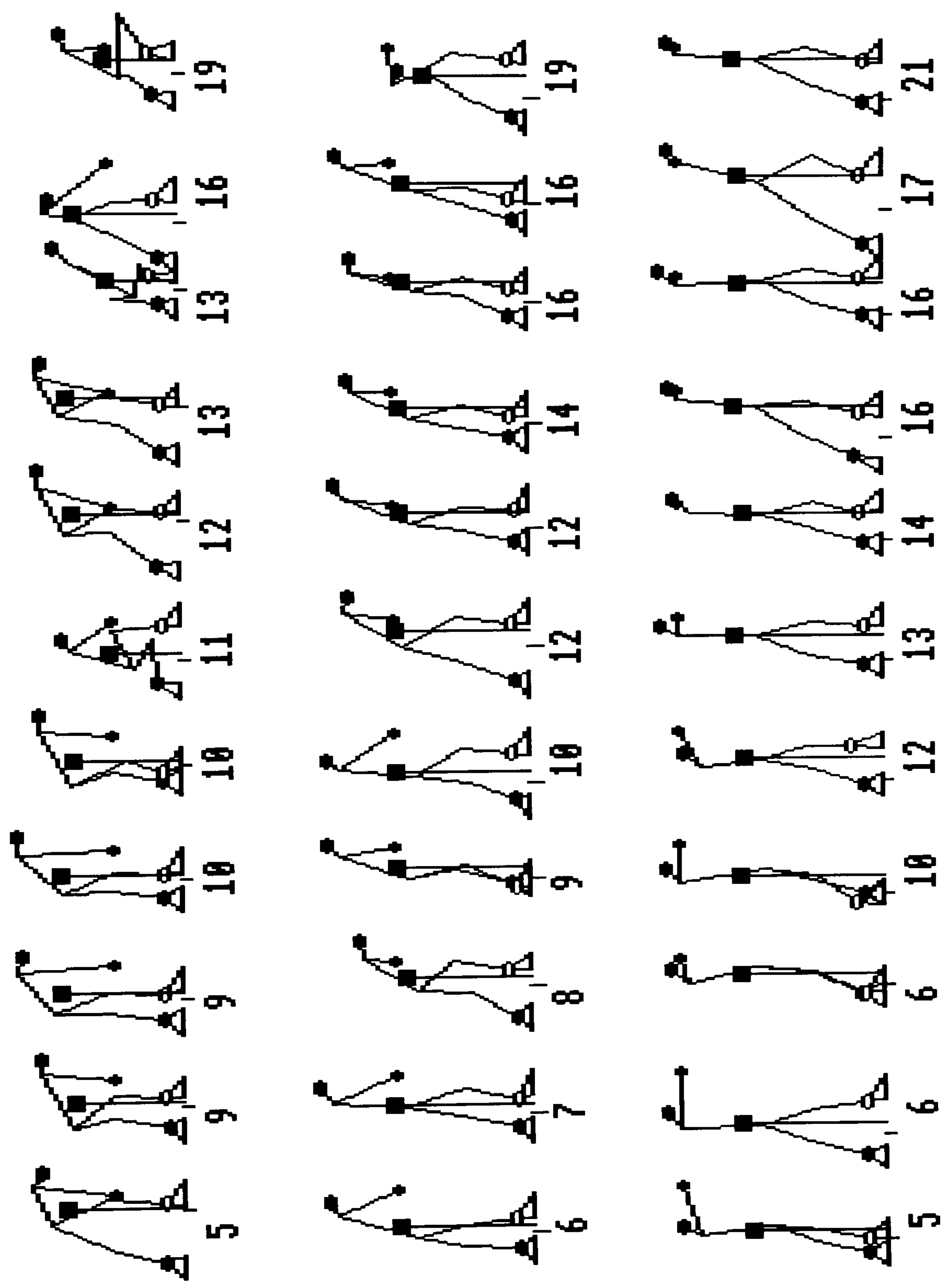
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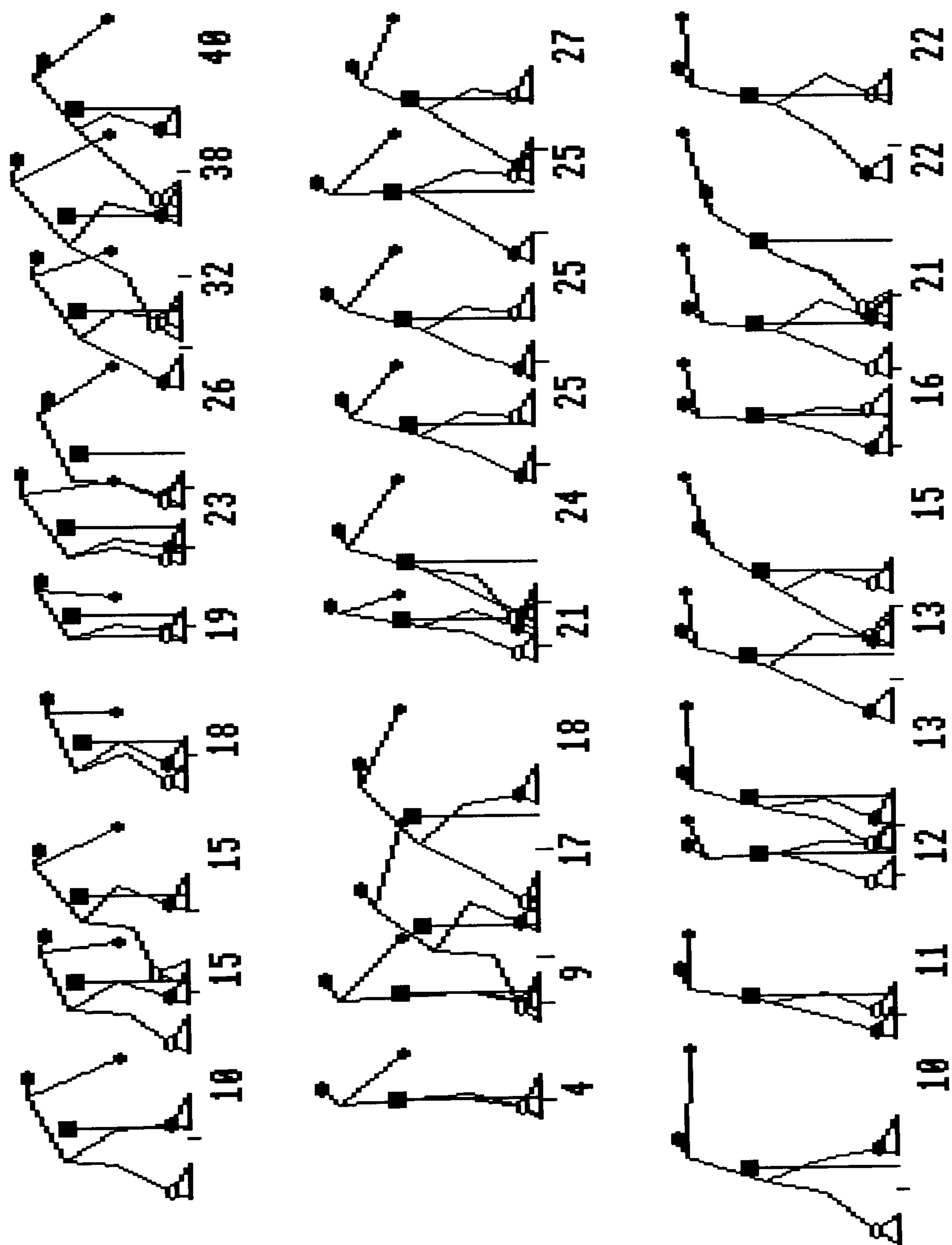
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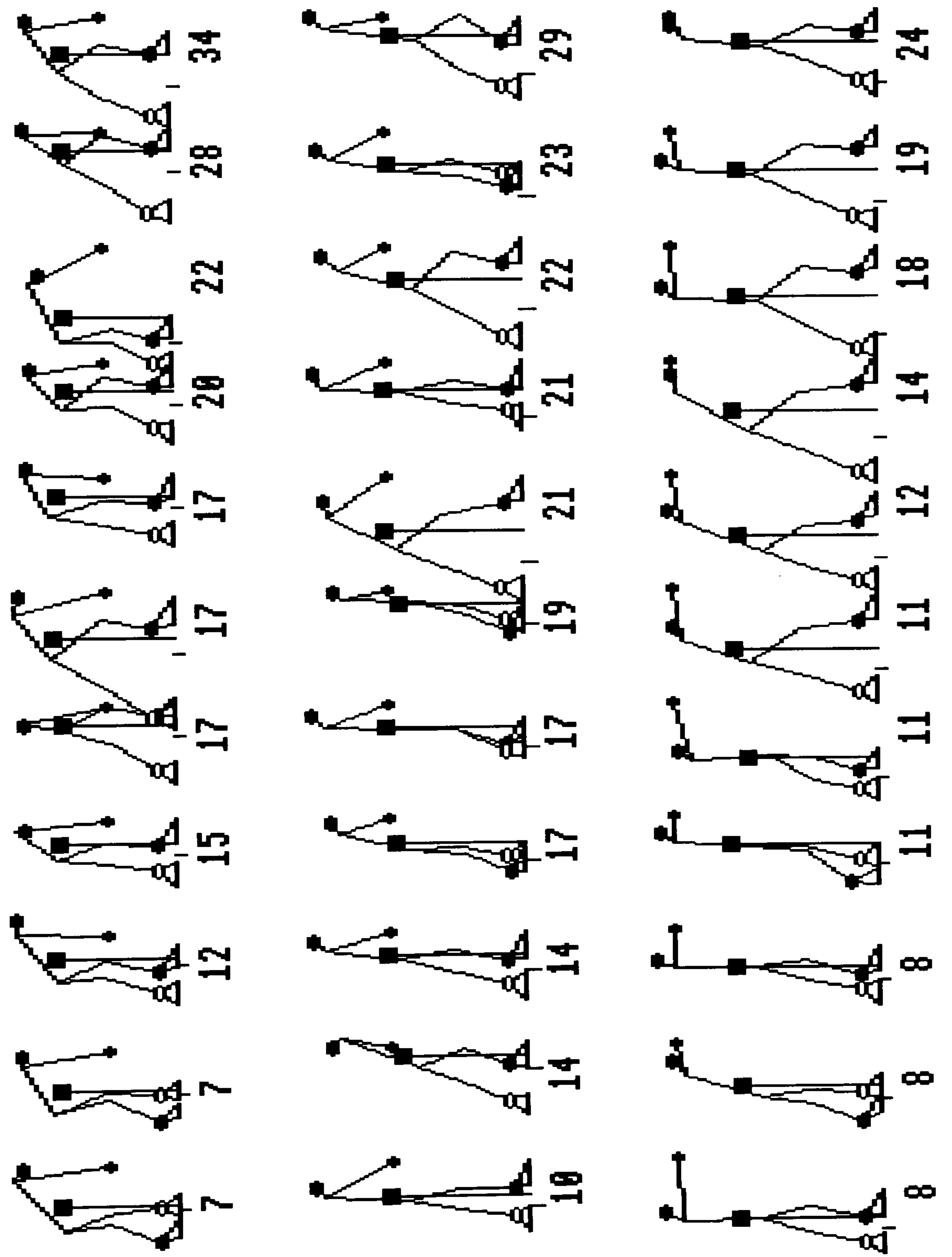
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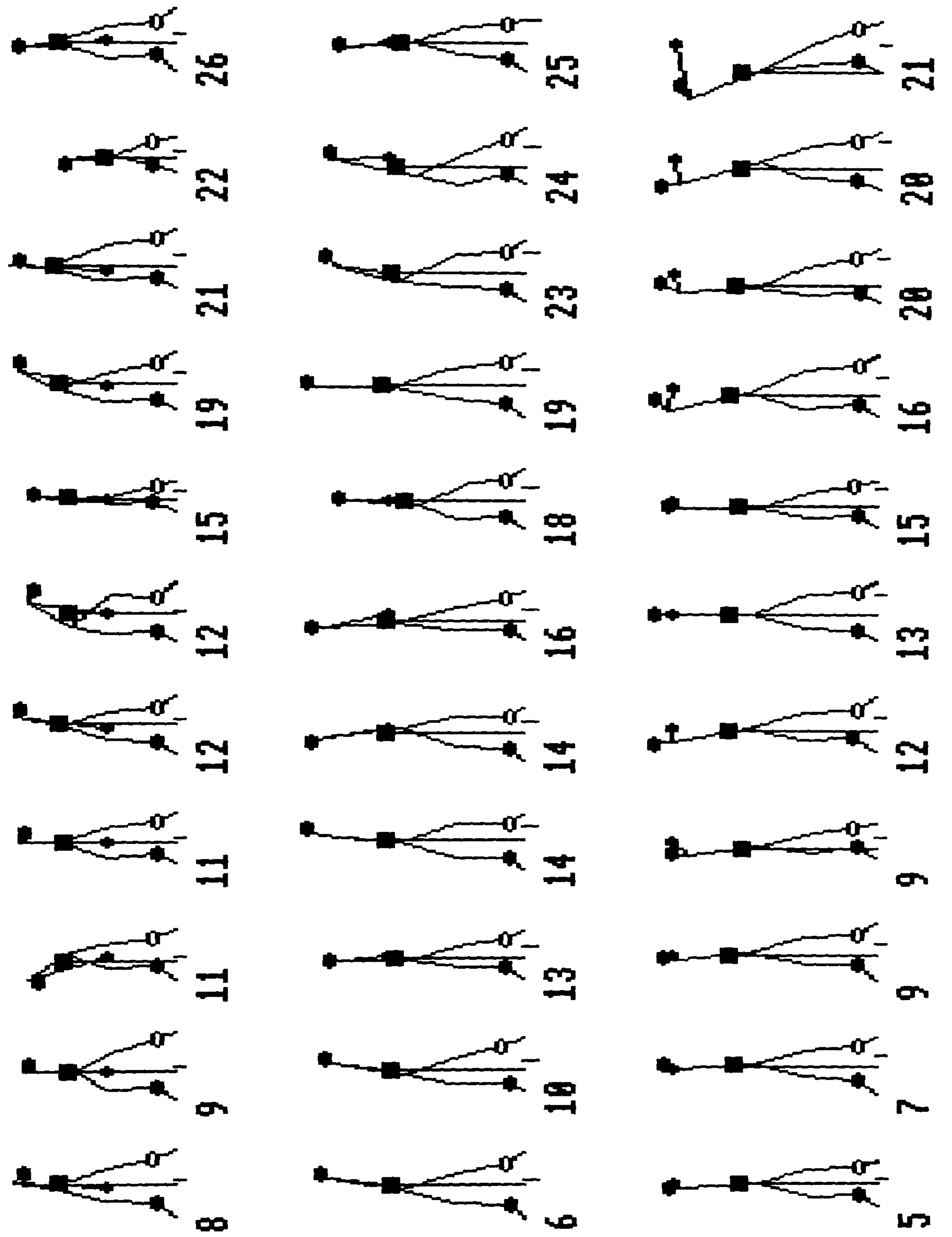
DIRECTION OF EXERTION: DOWNWARD AND FORWARD



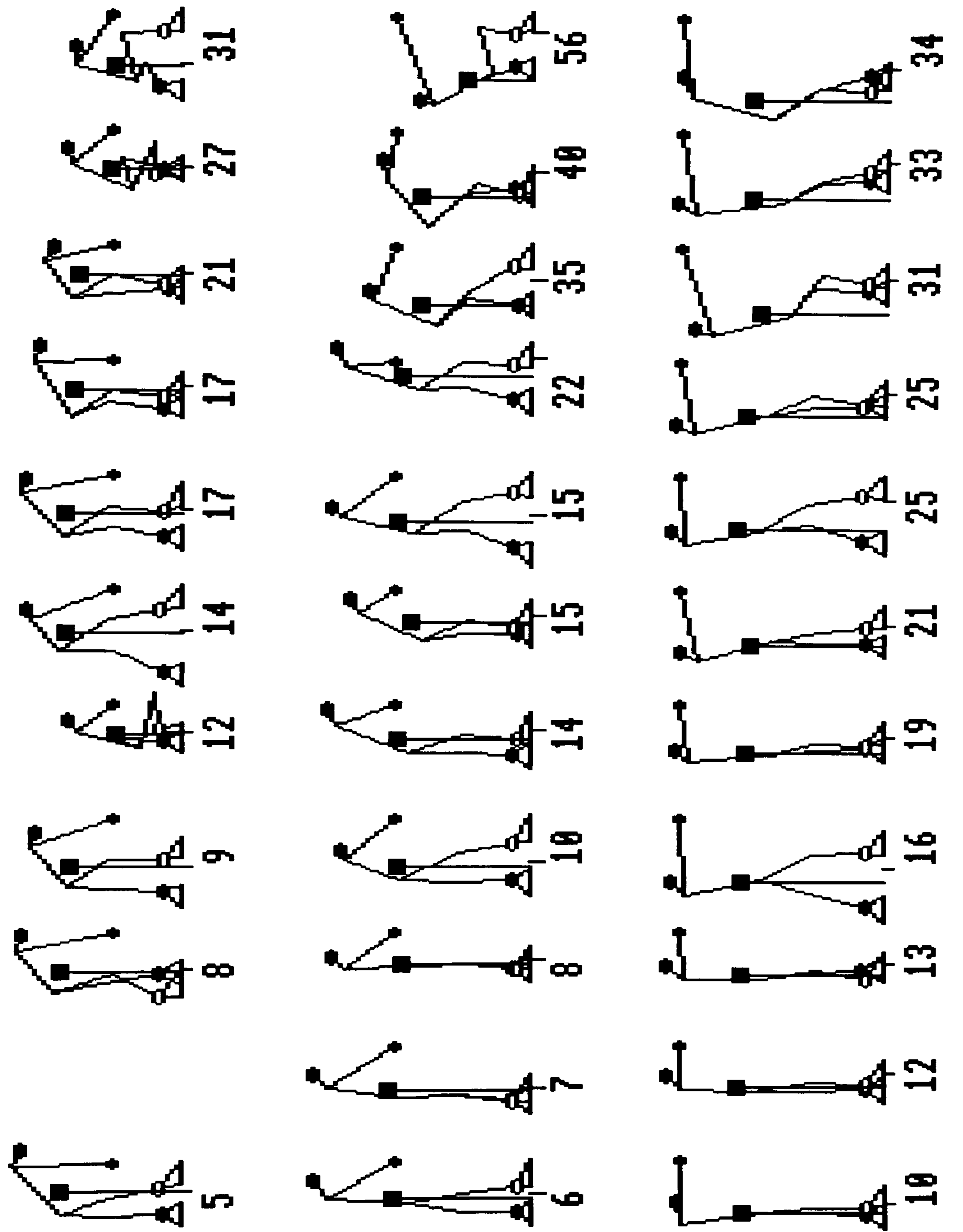
DIRECTION OF EXERTION: DOWNWARD, FORWARD & LEFT



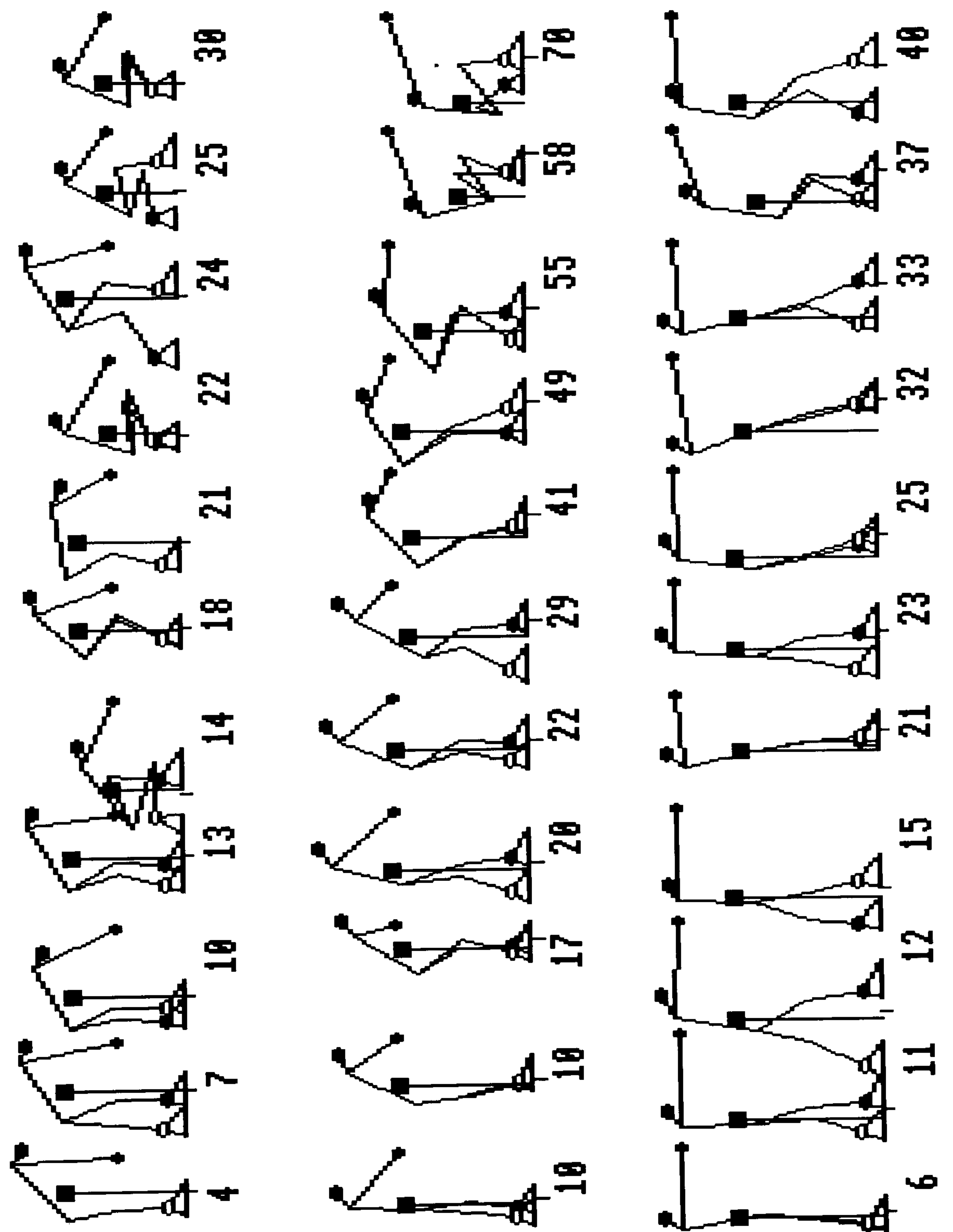
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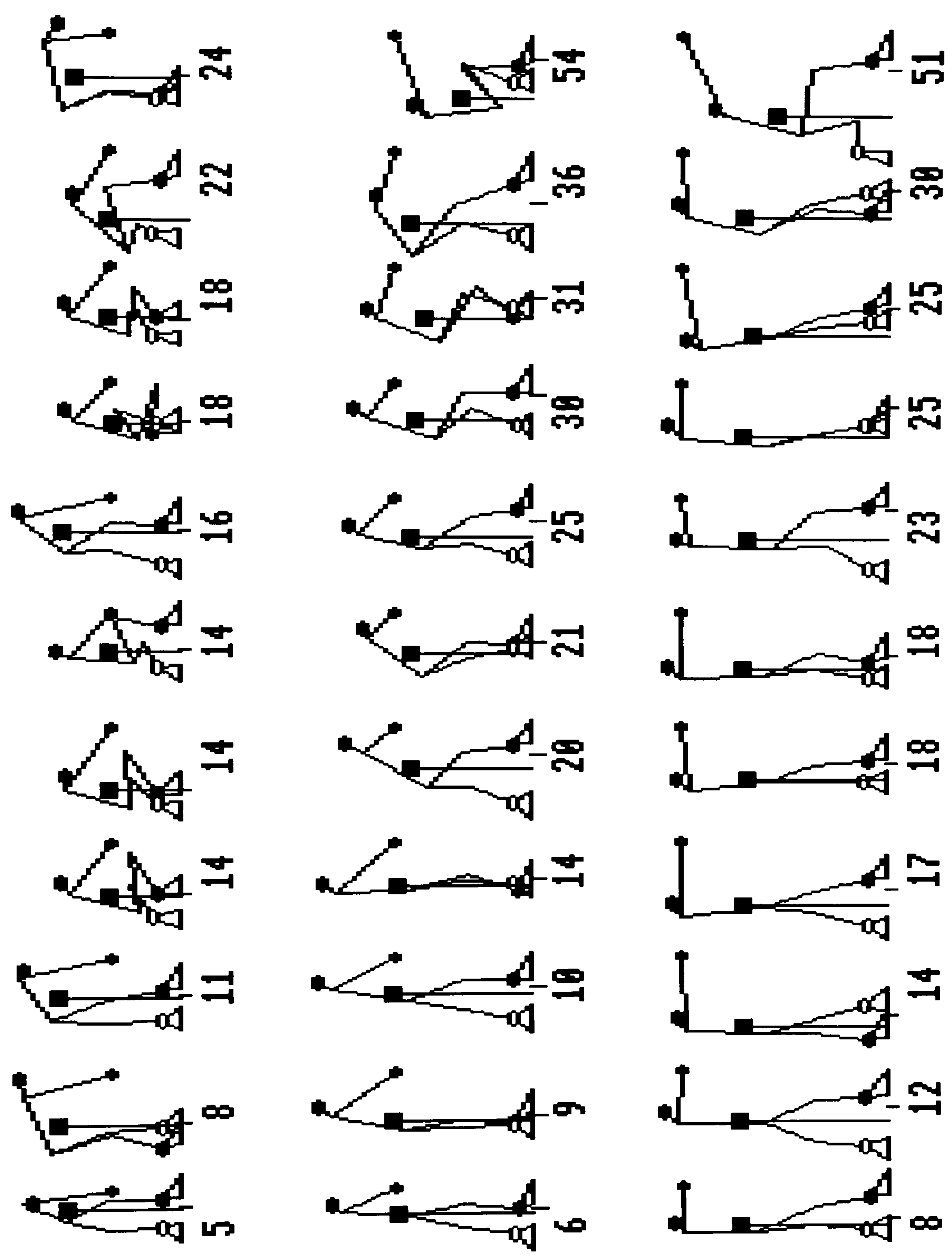
DIRECTION OF EXERTION: DOWNWARD, BACKWARD AND LEFT



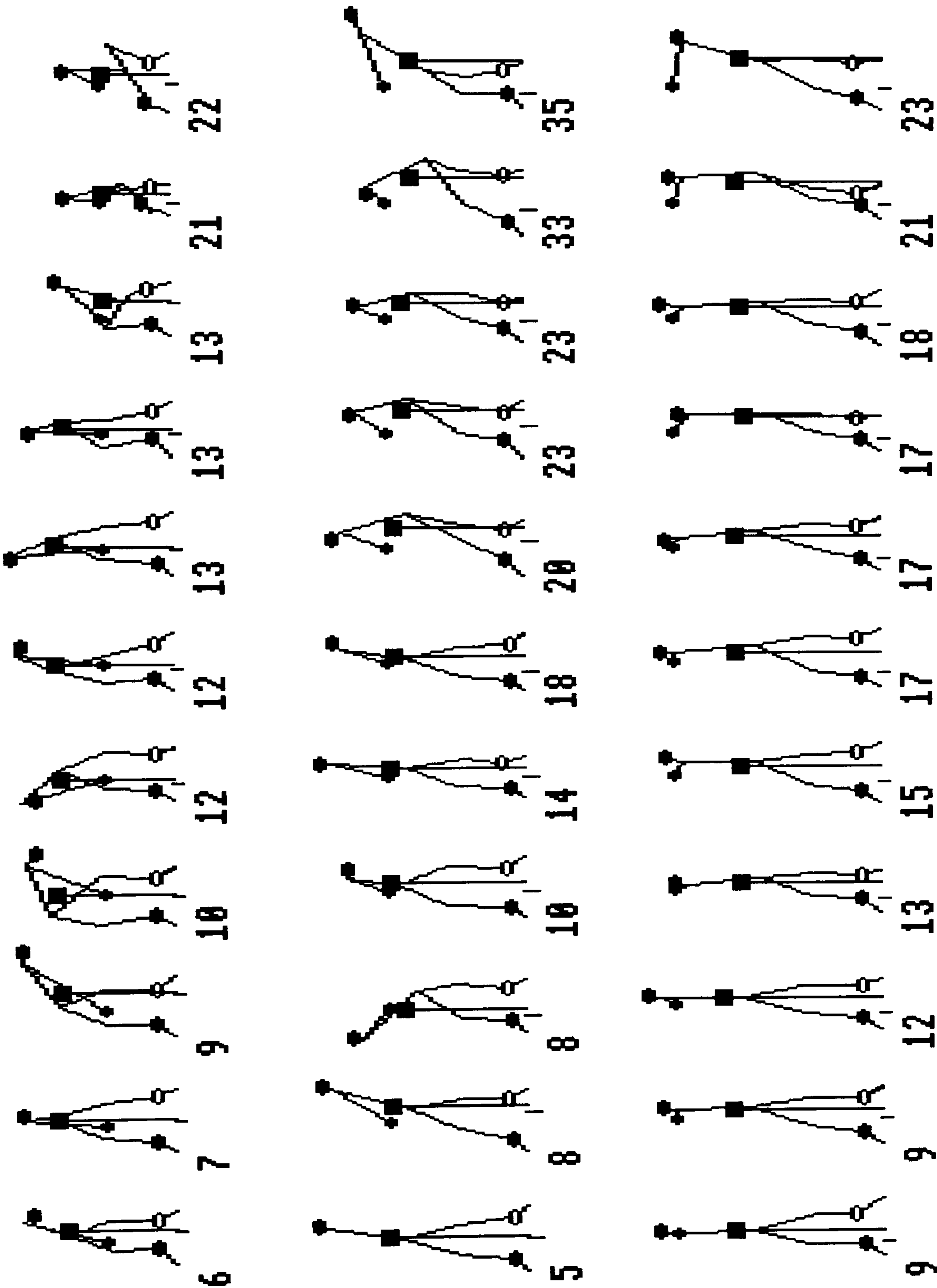
DIRECTION OF EXERTION: DOWNWARD AND BACKWARD



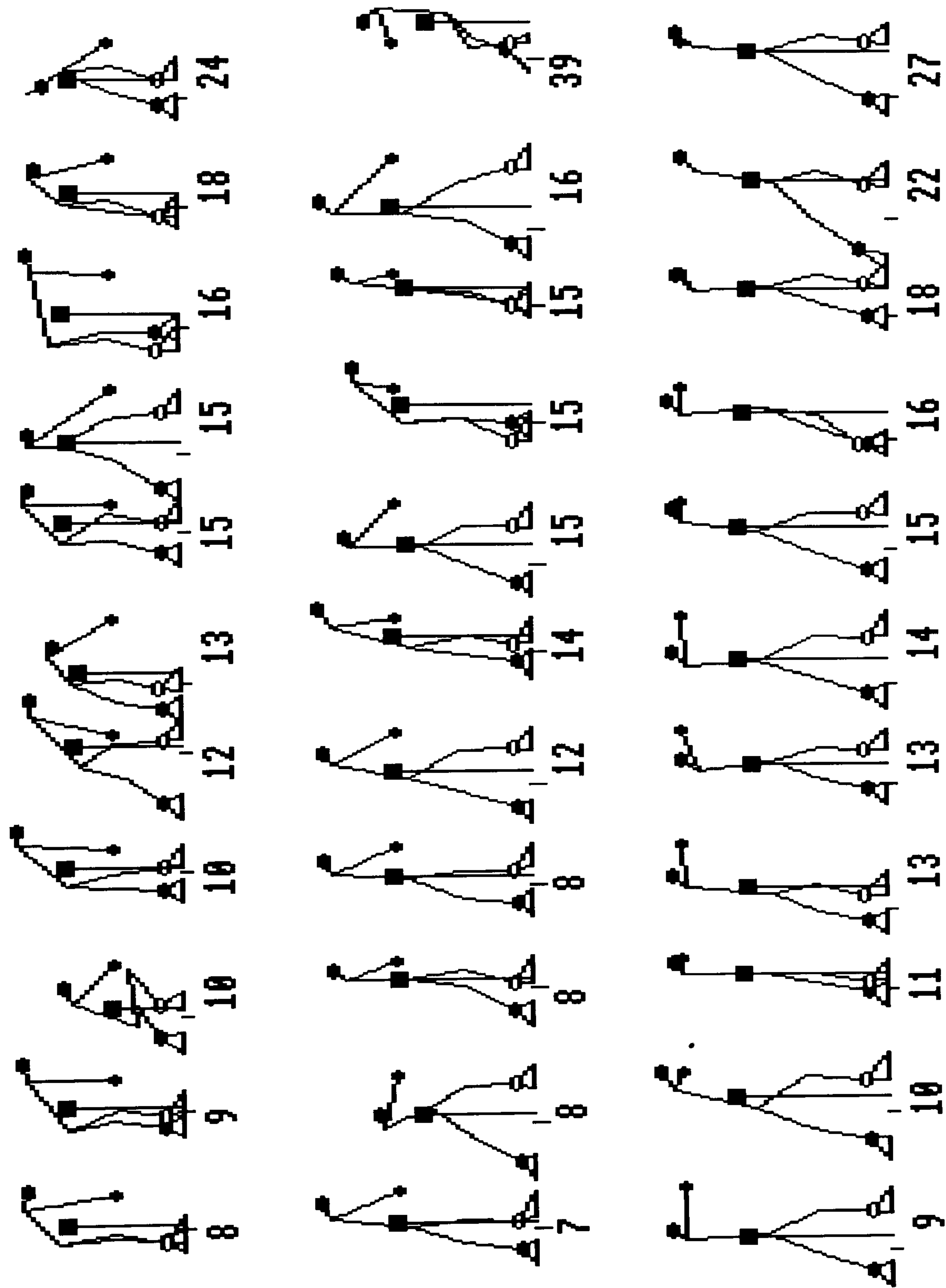
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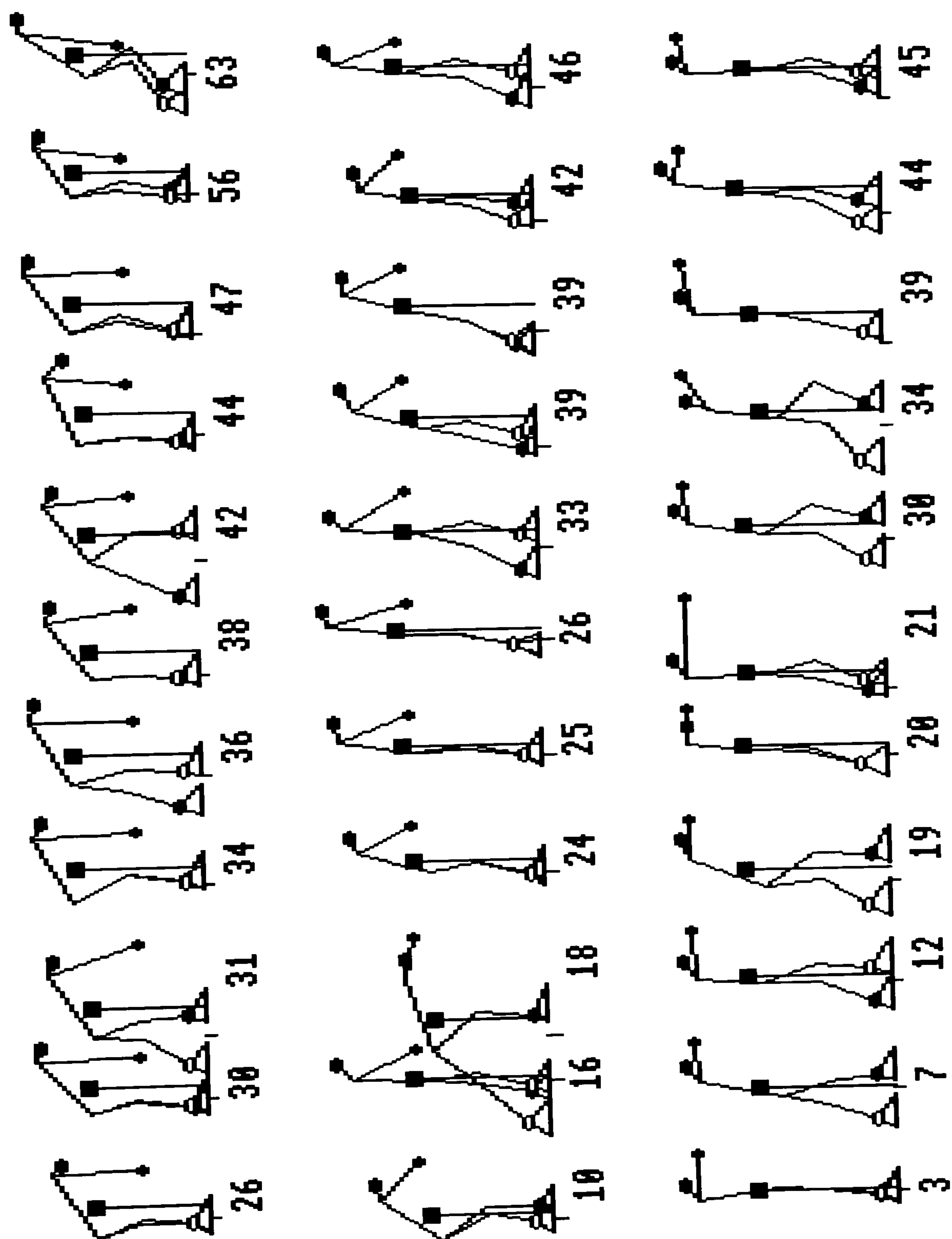
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DIRECTION OF EXERTION: DOWNWARD, FORWARD & RIGHT

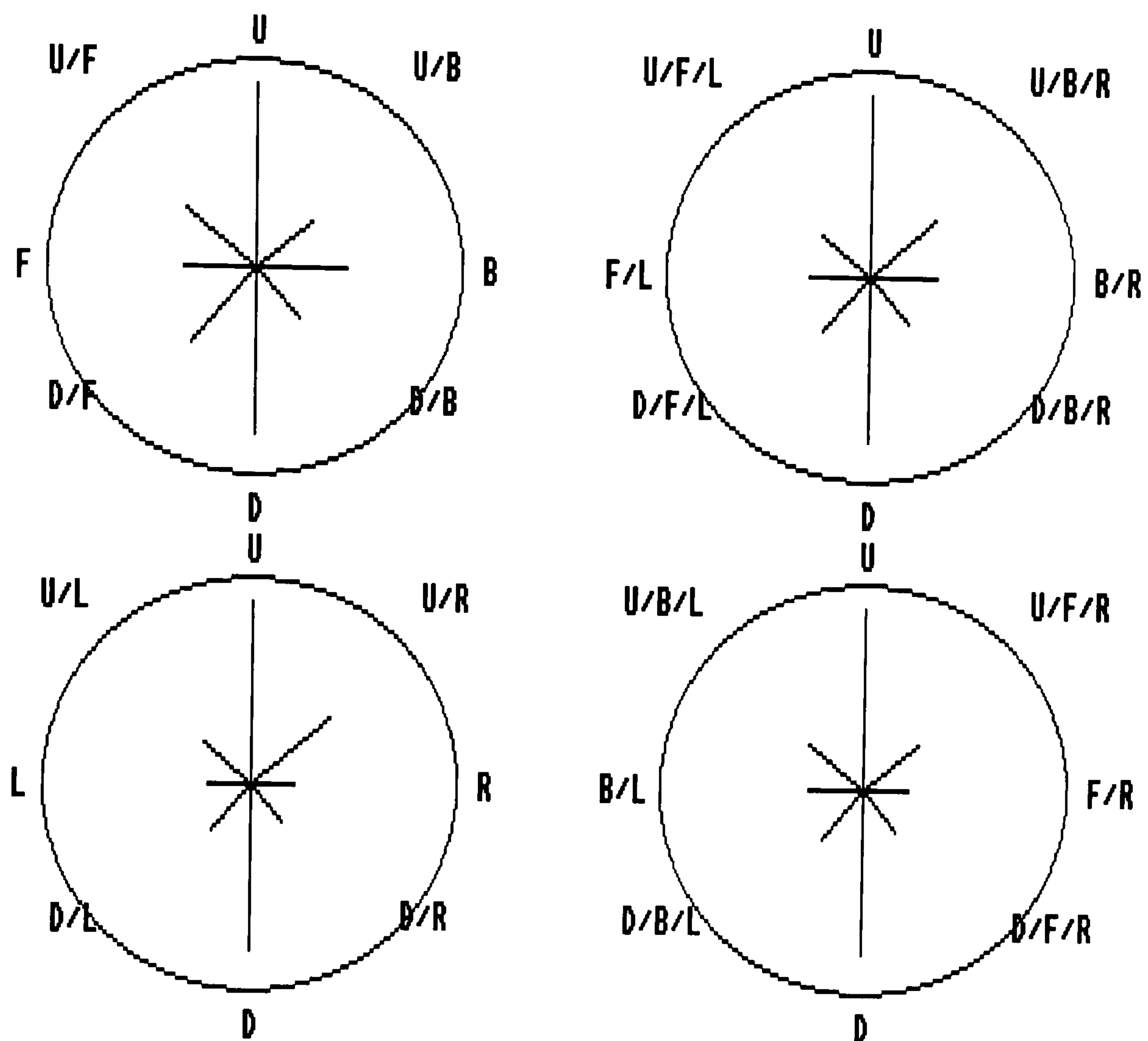


DIRECTION OF EXERTION: VERTICALLY DOWNWARD

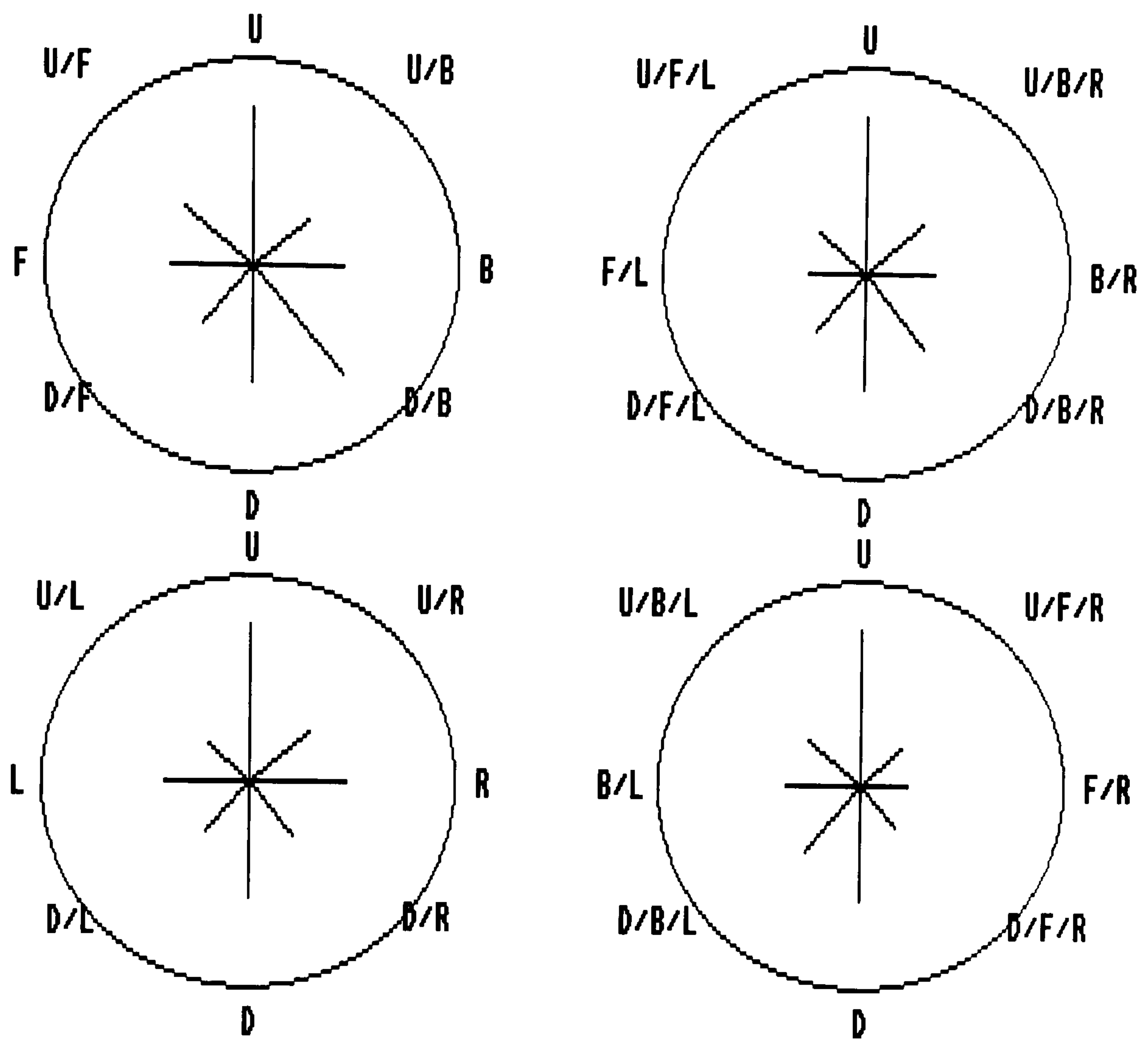


APPENDIX B.

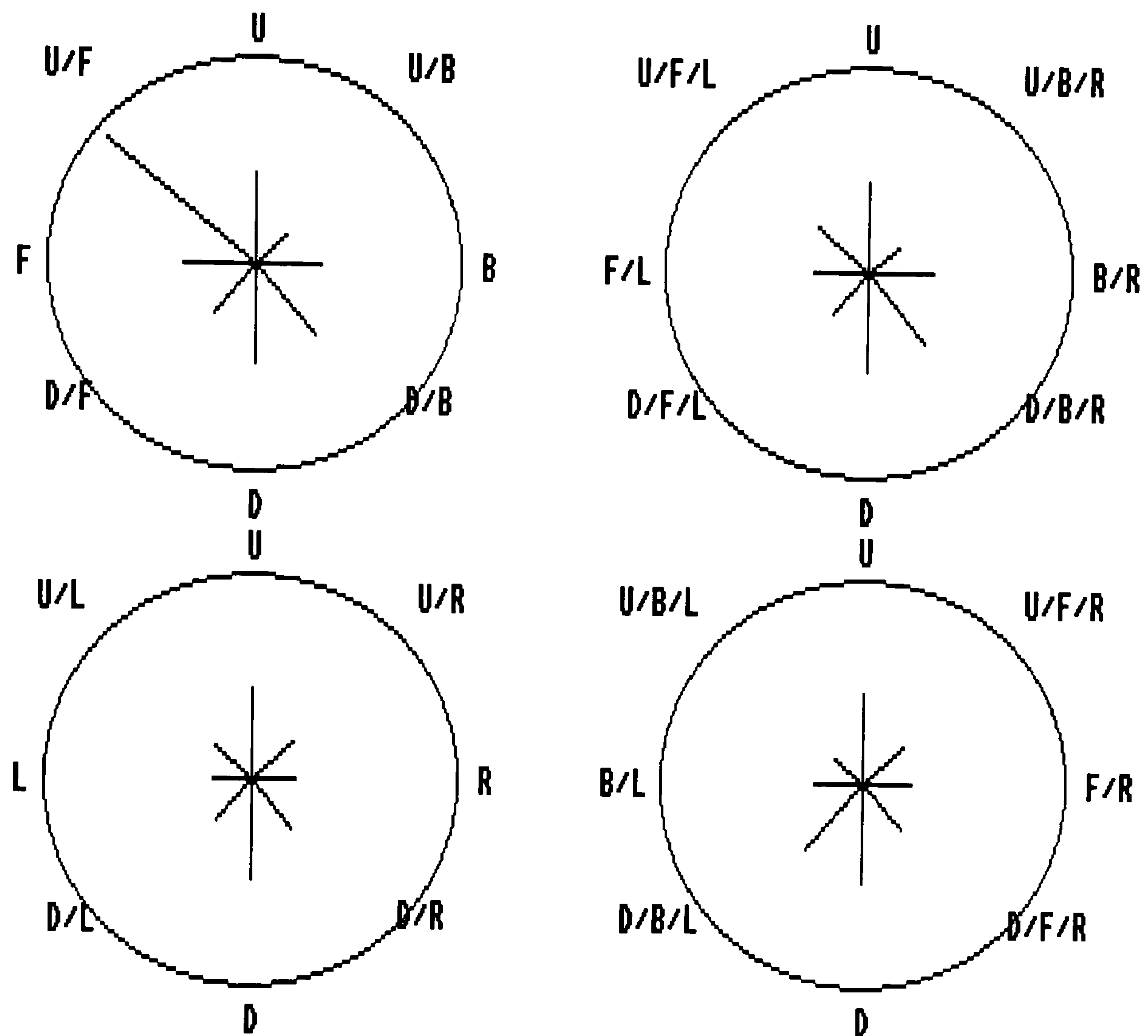
Forces exerted by subjects in Chapter 3.



Vertical slices showing mean magnitudes of forces (% body weight) exerted by subjects from Chapter 3 in all twenty-six directions at handle height 0.5m. The circles represent 100% body weight. U - up, D - down, F - forward, B - backward, L - left, R - right.



Vertical slices showing mean magnitudes of forces (% body weight) exerted by subjects from Chapter 3 in all twenty-six directions at handle height 1.0m. The circles represent 100% body weight. U - up, D - down, F - forward, B - backward, L - left, R - right.



Vertical slices showing mean magnitudes of forces (% body weight) exerted by subjects from Chapter 3 in all twenty-six directions at handle height 1.5m. The circles represent 100% body weight. U - up, D - down, F - forward, B - backward, L - left, R - right.

APPENDIX C.

Proportional anthropometry used in biomechanical model in Chapter 3.

Body segment	Position of centre of gravity	% body weight
Head and Neck	Anterior to tragus, superior to mandible	8.4
Trunk	54% from C7, 46% from pelvis	50.0
Thigh	41% from pelvis, 59% from knee	10.0
Leg	44% from knee, 56% from ankle	4.3
Foot	47% from ankle, 53% from big toe	1.4
Upper limb	as explained in Chapter 3 (see section 3.2.3, p.86 and Figure 3.8, p.87)	5.1

Table showing positions of the centres of gravity and the weights of body segments used in biomechanical model in Chapter 3 (from Pheasant, 1986, Dempster 1955).

APPENDIX D.

Subject anthropometry from all experiments.

Subject	Age (yrs)	Height (m)	Weight (kg)
1	22	1.76	75.2
2	28	1.89	91.6
3	28	1.79	84.2
4	29	1.75	74.0
5	26	1.78	66.1
6	31	1.78	65.8
7	31	1.63	69.5
8	23	1.75	87.3
9	30	1.75	81.6
10	24	1.68	70.9
11	22	1.72	59.4
Mean	26.7	1.75	75.0
Std Dev	3.3	0.06	9.6

Subject anthropometry for experiment in Chapter 3.

Subject	Age (yrs)	Height (m)	Weight (kg)
1	30	1.75	81
2	26	1.71	79
3	27	1.73	87
4	22	1.77	53
5	34	1.64	65
6	20	1.80	79
7	32	1.83	81
8	24	1.79	67
9	33	1.86	78
10	32	1.87	89
Mean	28.0	1.78	75.9
Std Dev	4.7	0.66	10.5

Subject anthropometry for surface electromyography experiment in Chapter 4.

Subjects	Age (yrs)	Height (m)	Weight (kg)
1	22	1.77	54
2	30	1.75	81
3	26	1.70	78
4	24	1.79	69
5	26	1.93	106
6	37	1.74	65
7	25	1.70	65
8	27	1.68	65
9	34	1.64	64
10	20	1.80	73
11	32	1.85	80
12	36	1.67	95
13	24	1.72	69
14	33	1.87	95
Mean	28.3	1.76	75.6
Std Dev	5.2	0.08	14.0

Subject anthropometry for force plate experiment in Chapter 4.

APPENDIX E.

Related publications.

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PUBLICATIONS.

Papers:

A.T.Wilkinson, A.D.J.Pinder, D.W.Grieve.

Relationships between one-handed force exertions in all directions and their associated postures. *Clinical Biomechanics* 10(1), 21-28, 1995.

A.D.J.Pinder, A.T.Wilkinson, D.W.Grieve.

Omnidirectional assessment of one-handed manual strength at three handle heights.

Clinical Biomechanics 10(2), 59-66, 1995.

Abstracts:

A.T.Wilkinson, A.D.J.Pinder, D.W.Grieve.

Postural aspects of three-dimensional exertion. *Proceedings of the XIVth congress of the International Society of Biomechanics* pp 1462-63, 1993.

A.D.J.Pinder, A.T.Wilkinson, D.W.Grieve.

Omni-directional assessment of one-handed strength. *Proceedings of the XIVth congress of the International Society of Biomechanics* pp 1046-47, 1993.